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University of Alberta

Source Modeling, Detection, and Processing

of the Electrooculogram Signal for

Control of a Mobile Ocular Prosthesis

by



Cheryl Louise Card

A thesis submitted to the Faculty of Graduate Studies and Research in
partial fulfillment of the
requirements for the degree of Master of Science

Department of Electrical and Computer Engineering

Edmonton, Alberta

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University of Alberta
Faculty of Graduate Studies and Research

The undersigned certify that they have read, and recommend to the Faculty of Graduate Studies and Research for acceptance, a thesis entitled "Source Modeling, Detection, and Processing of the Electrooculogram Signal for Control of a Mobile Ocular Prosthesis" submitted by Cheryl Louise Card in partial fulfillment of the requirements for the degree of Master of Science.



Abstract

An ocular prosthesis that provides natural-looking movement of the artificial eye is being developed at the University of Alberta. This thesis describes work done to determine how to extract a control signal from the natural eye movements, to drive the movements of the ocular prosthesis. Different methods of detecting eye movements are discussed, and the electrooculogram (EOG) method is suggested for the ocular prosthesis system. The results of EOG signal processing experiments are described, and a 5th-order Chebyshev type-II filter is recommended for de-noising the EOG signal.

A finite element model of the EOG source was developed. An anatomically accurate head model and a distributed biopotential source were used. Polynomial pre-conditioning was used to improve algorithm performance. The surface potentials predicted by the EOG source model were compared with recorded surface potentials to verify the model. The model was then used to determine electrode placement in the ocular prosthesis system.

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Glossary

<i>canthi</i>	The corners of the eye opening (palpebral fissure)
<i>cornea</i>	Transparent, refractive medium located at the front of the eye
<i>coronal plane</i>	A longitudinal plane dividing the body into front and back parts (also known as the frontal plane)
<i>dermis</i>	The layers of skin beneath to the epidermis
<i>epidermis</i>	The layers of skin nearest the skin surface
<i>exenterated orbit</i>	An orbit from which the eyeball, extraocular muscles, and supporting tissues have been removed
<i>fovea</i>	The region of the retina having the highest visual accuity
<i>intraorbital</i>	Contained within the orbit
<i>iris</i>	The pigmented ring surrounding the pupil of the eye
<i>ocular</i>	Pertaining to the eye
<i>optic disc</i>	The region of the retina at which the optic nerve enters the eye
<i>orbit</i>	The cavity in which the eyeball is contained
<i>retina</i>	The interior rear (dorsal) surface of the eyeball that contains light-sensitive cells
<i>saccade</i>	Fast, voluntary eye movement that brings an object of interest to the fovea of the retina
<i>transverse plane</i>	A longitudinal plane dividing the body into top and bottom parts
<i>sclera</i>	The unpigmented area of the eye adjacent to the iris

<i>subcutaneous</i>	Beneath the dermal layers of the skin
<i>frontal sinus</i>	Air-filled space in the frontal bone just above the eyes
<i>sphenoid sinus</i>	Air-filled space in the sphenoid bone behind the nasal cavity
<i>ethmoid sinus</i>	Air-filled spaces in the nose between the eyes
<i>maxillary sinus</i>	Air-filled spaces in the maxilla bones near the orbital cavities

Chapter 1

Introduction

The loss of an eye due to injury or disease can result in significant disfigurement. Fortunately, realistic-looking ocular prostheses are currently available to patients. However, the presence of the prosthesis becomes obvious when the patient moves his or her natural eye because the prosthesis does not move. This is a problem in day-to-day life because the immobile prosthetic eye is distracting to the observer. Patients can suffer significant psychological trauma from the unwanted attention the prosthesis elicits, and as a result some prosthesis wearers avoid normal social interaction [1]. An ocular prosthesis that moved in a natural-looking manner could provide patients with a less distracting appliance, thereby improving their ability to interact with others as they did before the loss of the eye. Such a prosthesis is in demand by both patients and health care professionals [2].

In order to determine whether or not an ocular prosthesis that provides natural-looking eye movement is feasible, a working bench-scale model of an ocular prosthesis that tracks the horizontal movements of the natural eye was developed by a group of researchers in Edmonton, Alberta [3] [4]. The group consisted of members of the Craniofacial Osseointegration and Maxillofacial Prosthesis Rehabilitation Unit (COMPRU) of the Misericordia Hospital, the Faculty of Rehabilitation Medicine, the Department of Industrial Design, the Department of Mechanical Engineering, and the Department of Electrical and Computer Engineering, all of the University of Alberta. This initial model of the prosthetic eye proved that the development of

such a prosthesis is possible, and paved the way for the development of a clinical model of the ocular prosthesis.

1.1 The Ocular Prosthesis System

The population who could benefit from the proposed ocular prosthesis system are those patients who have lost an eye and its surrounding structures to injury or disease. In these cases, the orbit is exenterated, removing all tissue associated with the eye. A flap of skin is attached across the back of the orbital cavity, covering the bony orbit.

In order to allow the ocular prosthesis system to track the movements of the natural eye, it is necessary to detect natural eye movement. In Chapter 2, it is suggested that this should be accomplished by using electrodes to detect the electrooculogram of the natural eye. The EOG is the signal sensed at the skin surface due to the ocular standing potential, and is proportional to the position of the natural eye. The signal recorded by the electrodes will be proportional to the movements of the natural eye.

The clinical ocular prosthesis system will use electrodes that are implanted subcutaneously near the natural eye. The electrooculogram signal detected by the electrodes will then be amplified by custom-designed circuitry implanted near the electrodes, and then transmitted to the ocular prosthesis located in the exenterated orbit by means of a thin wire. This wire will pass under the skin of the forehead and will enter the prosthesis through one of the orbital bones. This design will be used so that the presence of these components of the prosthesis will not be obvious. The feasibility of this design has been verified by the surgeons who will perform the

implantation procedure [2].

Once the electrooculogram signal has been received by the ocular prosthesis, the signal will be digitized and processed to determine the position of the natural eye. This information will then be used to drive the ocular prosthesis to track the movements of the natural eye. The prosthesis will be driven by an actuator that permits the prosthesis to rotate about its vertical axis, mimicing the (horizontal) movements of the natural eye. The ocular prosthesis system is illustrated in Figure 1.1.

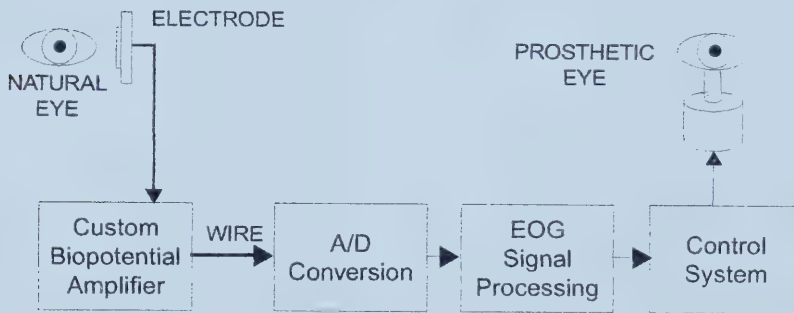


Figure 1.1: Ocular Prosthesis System

The diameter of the exenterated orbit is approximately 25 mm, and the volume of the cavity is similar to the volume of a ping-pong ball [2]. The entire ocular prosthesis must be contained within this volume, so that the presence of the prosthesis is not noticeable to the observer. The exenterated orbit must contain the cosmetic eye surface, the eye movement detection system, the actuator and its control circuitry, power management and storage facilities, and diagnostic functionality. The prosthesis must be removable for sanitary reasons and to recharge the battery, and must

attach to a fitting that is anchored to the patient's maxilla bone through the floor of the orbital cavity. Obviously, this project poses significant technical challenges. In addition to the electrical and mechanical design issues, the components of the prosthesis must be miniaturized so that they fit into the available volume. Also, the prosthesis system must be constructed of biocompatible materials that can remain in the body for years at a time.

It is evident that the entire project is beyond the scope of a Masters thesis. Therefore, this thesis is limited to the detection and processing of eye movement data into a usable control signal. In order to determine the correct placement of the subcutaneously implanted electrodes, it is necessary to simulate the electrooculogram signal at the potential electrode sites. For this purpose, a model of the source of the electrooculogram was developed, and is described in this thesis. The model was based on anatomical and physiological evidence of the source of the electrooculogram, and was verified by comparison with experimentally obtained data. The model was then used to determine the optimum placement for the subcutaneous electrodes. Signal processing methods for extracting a control signal from the raw eye movement data are also presented. In choosing the methods used for the detection and processing tasks, care was taken to ensure that the proposed solutions were compatible with the constraints imposed by the nature of the ocular prosthesis system.

1.2 Design Specifications for Ocular Prosthesis System

The specifications for the ocular prosthesis system were developed in conjunction with the medical staff at the Craniofacial Osseointegration and Maxillofacial Pros-

thesis Rehabilitation Unit (COMPRU) of the Misericordia Hospital, Edmonton, Alberta, and members of the Department of Electrical and Computer Engineering, the Department of Mechanical Engineering, and the Faculty of Rehabilitation Medicine, all of the University of Alberta in Edmonton, Alberta.

The primary consideration in developing these specifications was that the ocular prosthesis system be a cosmetic improvement upon the currently available ocular prostheses, which look very realistic. To achieve this goal, the movements of the prosthetic eye must be indistinguishable from and synchronous with the horizontal eye movements of the patient's natural eye. Also, the prosthesis must be physically compatible with the patient's medical condition and lifestyle.

The secondary consideration was the physical and electrical limitations inherent in an intraorbital prosthesis. The volume of the actuator, circuitry, power source, and prosthesis enclosure were dictated by the size of the average adult human bony orbit (about 25 mm in diameter). It was assumed that the distance between the two conversing parties will be between 0.5 m and 2.0 m, depending on the situation and on the cultural background of the individuals. Naturally, it is only necessary to design the eye system to replicate those eye movements that are noticeable to the human observer at this distance. The design specifications for the ocular prosthesis system are given in Table 1.1.

The supply voltage and current specifications were based on the requirements of the electrical components that would be used in the ocular prosthesis system. The run time and recharge time were based on the normal patterns of prosthesis usage by patients.

The maximum angular displacement specification was determined from the max-

Specification	Value (Units)
Outside diameter of prosthesis	25 mm
Supply voltage	3 V
Average current	200 mA
Run time	8 h
Recharge time	8 h
Maximum angular displacement	$\pm 45^\circ$ arc
Maximum angular velocity	1000° arc/s
Maximum angular acceleration	$\pm 10,000^\circ$ arc/s ²
Resolution	$\pm 1.0^\circ$ arc
Rise time (impulse response)	50 ms
Steady-state ripple (step response)	± 10 min arc
Maximum ripple frequency (step response)	150 Hz

Table 1.1: Design Specifications for Ocular Prosthesis System

imum range of an eye movement. Similarly, the maximum angular velocity and acceleration were determined from the characteristics of a typical eye movement [5]. The resolution corresponds to the resolution of the EOG recording, which is ± 1.0 degrees of arc [6].

The impulse response rise time is one quarter of the time between a visual stimulus and the resulting eye movement [5]. The steady-state ripple magnitude and frequency are taken from the amplitude and frequency of microsaccadic eye movements, which are below the threshold of observable eye movements [3].

1.3 Structure of Thesis

This thesis is divided into six chapters. The different types of eye movements as well as methods for detecting these movements are described in Chapter 2. The use of the electrooculogram (EOG) to detect eye movements is proposed, and the issues

related to detecting the EOG in the clinical model of the prosthesis are discussed. Suitable methods for extracting a usable control signal from the EOG that will be detected in the clinical model are discussed in Chapter 3. Several signal processing methods for de-noising the EOG are presented, as are the results of signal processing experiments involving simulated and actual EOG data.

In order to determine the proper placement of the electrodes in the clinical ocular prosthesis system, a numerical model of the source of the EOG is developed in Chapter 4. Finite element modeling techniques are used to verify that the model is correct by comparing the surface potentials predicted by the model with actual measured data. The EOG source model is then used to determine the optimum electrode configuration for the clinical model of the prosthesis. These results are presented in Chapter 5. Chapter 6 contains the conclusions of this thesis.

1.4 Scientific Contributions of Thesis

The work described in this thesis contributes necessary technical information regarding signal acquisition and processing to the team that is developing the ocular prosthesis system. This information will facilitate the eventual development of a working clinical model of the prosthesis.

A contribution is also made to the body of knowledge concerning finite element modeling of biopotentials in the head, such as the electrooculogram and the electroencephalogram. This thesis uses a model that has a small element size and approximates the anatomy of the head, which is uncommon in the biopotential source modeling community. Most researchers use a more approximate head model due to

computational constraints, despite the fact that accuracy suffers as a result [7] [8] [9] [10]. The methods that were used to overcome the mathematical difficulties inherent in biopotential source modeling with an accurate head model are presented.

Also, the biopotential source model used in this thesis assumes a distributed source. Currently, most researchers assume that biopotentials in the brain arise from a single source. This is an oversimplification of the problem, and results in errors in source localization [11]. Therefore, distributed source modeling is of interest to the biopotential source modeling community.

The results of this thesis will also be of interest to the ophthalmology community [12]. The EOG is part of a battery of tests used in ophthalmology to assess the health of the retina. However, both the anatomical source of the EOG and the physiological mechanisms that give rise to this biopotential are poorly understood. A more accurate model of the source of the EOG would provide additional information about the causes and possible treatments of various disorders of the retina.

Chapter 2

Detection of Eye Movements

The different types of eye movements and their characteristics are discussed in this chapter. Various methods of sensing eye movements are presented, and the implications of each method for the design of the ocular prosthesis system are considered. Finally, the electrooculogram is recommended as the best method for sensing eye movements for the ocular prosthesis system, and the reasons for this recommendation are discussed.

2.1 Types of Eye Movements

Three pairs of antagonistic muscles control the movement of the eye. These are the lateral and medial rectus muscles, superior and inferior rectus muscles, and superior and inferior oblique muscles. These muscles are collectively known as the extraocular muscles, and are shown in Figure 2.1. Each muscle pair is primarily responsible for eye movement in one direction. The lateral and medial recti provide rotation about the vertical axis of the eye, the superior and inferior recti provide rotation about the horizontal axis, and the superior and inferior obliques provide rotation about the torsional (anterior-posterior) axis of the eye.

Several types of eye movements are executed by the extraocular muscles, including saccades, miniature eye movements, pursuit movements, and vergence movements [14]. Saccades are fast, voluntary movements that bring objects of interest to the

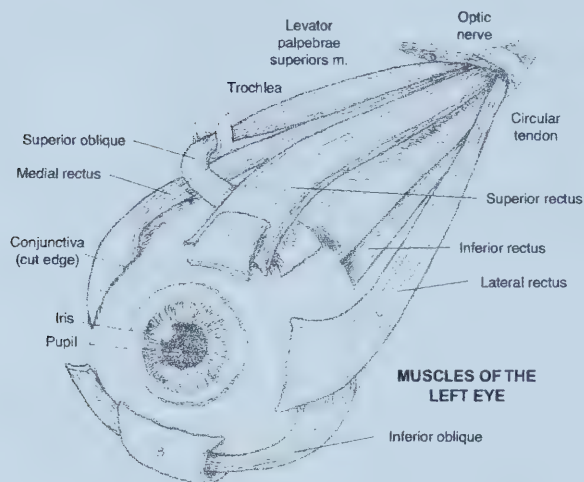


Figure 2.1: Extraocular Muscles [13]

fovea of the retina, which is the region of highest visual acuity. A familiar example of this type of eye movement is the scanning movements that occur when reading a page of text. In general, saccades result in an angular displacement of less than 15 degrees, an angular velocity of less than 1000 degrees/s, and an angular acceleration of less than 40 000 degrees/s². The duration of saccadic eye movements is between 20 ms and 100 ms, and the latency period between movements is 150 ms to 300 ms [5].

Miniature eye movements occur during an attempted steady gaze. These movements have a magnitude of less than 10 minutes of rotation and are involuntary

[3]. Pursuit movements are another type of eye movement, and include slow pursuit, smooth pursuit, and optokinetic movements. The slow and smooth types of pursuit movements occur when the head undergoes rotational acceleration. In this situation, the eyes rotate about their vertical axes in the direction opposite to the head's rotation, in order to maintain the position of the image on the retinas of the eyes. These pursuit movements are slow and are generally less than 7 degrees of rotation [3]. Large rotations of the head cause the eyes to exhibit nystagmus, where the eyes alternate between slow pursuit movements in the opposite direction of the head rotation and fast anti-compensatory saccades in the same direction as the head rotation. These movements bring the new visual field into focus.

If the entire visual field is moving the eyes will exhibit optokinetic nystagmus. The reader will likely have experienced this phenomenon when looking out the window of a fast-moving vehicle. The eyes attempt to perform slow-tracking movements to follow objects in the visual field as they move past, and then quickly saccade back to find a new point of interest. The maximum rate at which the eye can track moving objects in this manner is about 30 degrees/s [3].

Vergence movements are movements of the eyes in opposite directions that are used to fuse images of near or far objects. These movements will not be reproduced by the ocular prosthesis system, because the goal of the system is to provide the patient with an improvement in the cosmetic appearance of the prosthesis during normal social interaction, such as a conversation. Vergence movements are not as important to this outcome as the other types of movements listed above, since the object of fixation (another person) is unlikely to move very near or very far away from the individual during a conversation.

It is assumed that both the head of the individual and the visual field will be essentially stationary, therefore pursuit eye movements will not be tracked. Eye movements of less than 1 degree of rotation are not noticeable to the observer [3], so only eye movements with a magnitude of more than 1 degree of rotation will be tracked by the prosthesis. Miniature eye movements will not be tracked since they have a magnitude much less than 1 degree of rotation. Thus, the ocular prosthesis system will be required to track saccadic eye movements of magnitude greater than 1 degree of rotation. Saccadic eye movements are the most frequently occurring eye movements during normal social interaction, and as such are also the most important eye movements for the ocular prosthesis to reproduce in order to achieve a satisfactory cosmetic result.

2.2 Eye Movement Detection Methods

There are many methods of detecting eye movements. These include imaging the eye, optically sensing the iris-sclera boundary, magnetically sensing the position of the eye, and electrode methods [14].

Eye movements can be detected by applying image processing algorithms to video or still images of the eye [15]. This method requires that the imaging device be located in front of the eye, and that the device's position relative to the orbit remains constant. Executing the image processing algorithms consumes a large amount of computer processing power, and requires both space for the necessary electronics and sufficient available power.

Optically sensing the iris-sclera boundary can also be used to determine the

position of the eye [14]. This method uses infrared emitter and detector arrays positioned in front of the eye. Optical sensing also requires that the positions of the arrays remain constant with respect to the orbit. The position of such detectors cannot be guaranteed to be constant during normal prosthesis use. Therefore, the imaging and optical sensing methods were deemed to be unsuitable for the ocular prosthesis system.

The position of the eye can also be sensed magnetically [16]. This method requires that the individual wear a hard contact lens embedded with a magnetic coil on their natural eye. The electromagnetic field of the coil is then sensed to determine the position of the eye. This method would not be possible if the individual required corrective contact lenses. Also, the medical community is reluctant to introduce a foreign object into the one remaining natural eye of the individual for fear of the sight being lost in the remaining eye as well. For these reasons, the magnetic method is not suitable for sensing eye position in the ocular prosthesis application.

Electrode methods of sensing eye position measure the electromagnetic potentials that result from eye movements. One such method is the electrooculogram (EOG). An electrical potential exists in steady state between the cornea and the retina of the eye. This potential is known as the ocular standing potential. The EOG is the signal sensed at the skin surface due to the ocular standing potential, and is proportional to the position of the natural eye, as mentioned in Chapter 1. The EOG is typically recorded using silver/silver chloride bipolar electrodes situated at the canthi of the eye (the corners of the eye) and a forehead reference electrode [17]. This electrode configuration will sense horizontal eye movements. If the recording electrodes were placed above and below the eye instead of at the canthi, vertical

eye movements would be detected. The resolution of the EOG is approximately 1 degree, which satisfies the resolution specification given in Section 1.2 [18]. The EOG can be sensed non-invasively with external electrodes, and can also be sensed with subcutaneously implanted electrodes.

2.3 Detection Method for Ocular Prosthesis System

The EOG was chosen as the eye position detection method for the ocular prosthesis project. The reasons for this decision are that the EOG is easily accessible at the temple for both external and internal sensing, cross-coupling of the horizontal and vertical movements can be avoided by proper electrode placement, and the relationship between EOG signal and eye position is linear. Also, gross movement of the detection device with respect to orbit position is unlikely if the subcutaneously implanted EOG electrodes are applied properly. When the clinical model is implemented with subcutaneous electrodes, the electrooculogram method will provide the best compromise between being minimally invasive and maximally inconspicuous. As such, it will provide an excellent cosmetic result for the individual. Also, a control signal can be extracted from the EOG using the limited processing power of the ocular prosthesis.

The magnitude of the EOG signal is about 1 mV. To increase the signal strength, the EOG will be amplified with a custom-designed biopotential amplifier, which will be implanted under the patient's skin in the area of the sensing electrodes. The amplified signal will be transmitted to the prosthesis via a thin wire that passes under the skin of the individual's forehead. The amplified signal will be digitized

and processed to extract a control signal. This control signal will be the input to the system that governs the movements of the prosthetic eye.

Preliminary experiments conducted in the Advanced Robotics and Teleoperation Lab have shown that this design is capable of producing prosthesis movements that track natural eye movements, in the laboratory [4]. However, sensing the EOG in a clinical system presents several challenges that were not solved in the preliminary experiments. The EOG is corrupted with non-stationary noise from other biopotentials (the electrocardiogram, the electromyogram of the extraocular and facial muscles, and the electroencephalogram) and non-biopotentials such as 60 Hz line noise. This problem can be addressed by digitally processing the EOG signal, and is discussed in Chapter 3. Also, the proper position for the subcutaneously implanted electrodes must be determined; this issue is resolved in Chapters 4 and 5.

Chapter 3

Signal Processing

The EOG signal obtained in the ocular prosthesis system must be processed to extract the position of the natural eye, which will then be used as a control signal for driving the movements of the prosthesis. This chapter presents the characteristics of the EOG signal and discusses processing techniques that allow a control signal to be extracted from the EOG. Section 3.4 surveys the various methods that have been used in the past to process the EOG signal and evaluates the suitability of these methods for this application. In Section 3.5, a suitable signal processing method for the ocular prosthesis system is suggested, taking into account the constraints imposed by the limited processing power available in the clinical implementation of the ocular prosthesis.

3.1 Characteristics of EOG Signal

The EOG is the signal sensed at the skin surface that is due to the ocular standing potential. The EOG is typically recorded using a bipolar electrode montage, where one electrode is located at each canthus of the eye (the corners of the eye), as mentioned in Section 2.2. The EOG has a magnitude of less than 1 mV and a frequency less than 30 Hz [17]. The signal detected by an electrode is comprised of several components. These are the signal of interest (the EOG), undesirable biopotentials, 60 Hz line noise and its associated harmonics, and background noise. The undesir-

able biopotentials include the electroretinogram, the electromyogram from the ocular muscles, the electroencephalogram from the brain, and the electrocardiogram from the heart. These signals range between 0.1 mV and 10 mV and have a frequency between 1 Hz and 1000 Hz [17].

The EOG signal detected by an electrode must be amplified before signal processing techniques are applied to ensure that the signal of interest is strong enough to be detected by the digital signal processing circuitry. This is accomplished using a biopotential amplifier. Biopotential amplifiers are specialized amplifiers that address the problems encountered when measuring biopotentials such as the EOG. These amplifiers must amplify the signal of interest and reject noise and interference as much as possible. They also must protect the patient by providing isolation from the 60 Hz power grid and from leakage currents due to ground loops.

A typical biopotential amplifier involves several stages, as shown in Figure 3.1. These are a differential pre-amplifier, a high-pass filter to reduce DC and low frequency interference and noise, an isolation amplifier to provide protection against shocks, and a low-pass filter to reduce interference.

The pre-amplifier is designed to have a high common mode rejection ratio and a high input impedance. This design allows the amplifier to reject background noise and interference, since the EOG is a bipolar recording. To reject common mode interference, the common mode rejection ratio should be at least 100 dB. The input impedance of the amplifier should be much greater than the source impedance of the electrode (i.e. at least 10 M Ω) in order to avoid loading the signal of interest, which would distort the signal and degrade the performance of the amplifier. In order for the amplified signal to be compatible with digital hardware, the amplifier should also

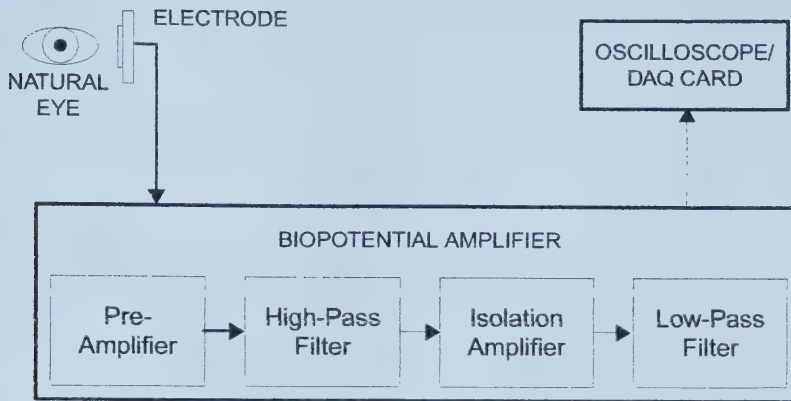


Figure 3.1: Biopotential Amplifier

provide a gain of at least 1000 with a very high signal to noise ratio [19].

The isolation amplifier should provide complete separation of the subject and the instrumentation by preventing current flow across the barrier, in order to protect the subject. This can be accomplished by using an optically coupled isolation amplifier. Care should also be taken to ensure that the subject is not part of a ground loop by connecting the patient to only one ground reference, so as to prevent shocks.

Electric and magnetic field interference can be reduced by shielding the electrode leads and grounding the leads at the amplifier, and also by twisting the leads around each other to reduce the effective area of any current loops.

A Grass 15RXi physiodata amplifier was used for the collection of EOG data in this project. This amplifier meets all of the criteria for a biopotential amplifier outlined above. The details of the data collection apparatus are discussed in the next section. For the clinical implementation of the ocular prosthesis, a custom

biopotential amplifier will be designed that satisfies the low power, small package requirements of this application.

3.2 Collection of EOG Data

3.2.1 Apparatus

The apparatus for collecting the EOG data consisted of electrodes, a biopotential amplifier, data acquisition hardware, and analysis software. The electrodes were Grass 6 mm reusable silver/silver chloride EEG electrodes. The skin surface was prepared for electrode application by cleaning the skin with isopropyl alcohol. The electrodes were filled with an electrolyte gel prior to application, and were affixed to the prepared skin using tape. The recording electrodes were placed at the temples. The reference electrode was placed on the chin, a location that is indifferent to the EOG signal.

The biopotential amplifier was a Grass RX15i physiodata amplifier system. This amplifier accommodates 16 channels of bipolar or referential analog input, and is equipped with a medical-grade isolation transformer for patient protection. Each channel has a low-pass and a high-pass filter that is configurable over a wide range of frequencies. Ten channels were used for this application. Each channel was configured to have a high-pass cut-off frequency of 0.01 Hz and a low-pass cut-off frequency of 30 Hz. These settings were used to preserve as much of the EOG signal as possible while attenuating other biopotentials and non-biopotential noise sources such as 60 Hz line noise. The gain of each channel was set to 10,000.

The data acquisition card was a National Instruments AT-MIO-64E-3. This card

can acquire data from up to 64 channels at a sampling rate between 1 kHz and 500 kHz, and has a maximum system noise of 0.15 LSB. For this application, 10 data acquisition channels were used, and the sampling rate was set to 1 kHz. LabView was used to acquire data from the AT-MIO-64E-3, and MatLab was used for data processing and analysis.

3.2.2 Ethical Approval

Ethical approval for the collection of data from human subjects for this project was obtained from the University of Alberta Health Research Ethics Board. In accordance with the requirements of the approval, each experimental subject was apprised of the nature of the experiment, its purpose, and the benefits and risks associated with participating in the study. Each subject signed a consent form before participating in the study. The file number for the ethical approval is B-120602-ENG. A copy of the approval is included in Appendix A.

3.2.3 Data Collection Methods

The experimental subject was fitted with recording electrodes as described above, and was allowed to adapt to the ambient light in the room for 10 minutes. This allowed the EOG to stabilize to within approximately $\pm 5\%$ of its steady-state value, in case there was any change in the EOG due to a change in illumination level when the subject entered the room [20]. The surface potentials were then recorded while the subject attempted to gaze straight ahead. Five trials of 10 seconds each were performed and recorded, with as much time between trials as the subject felt was necessary.

3.3 Signal Acquisition

The EOG signal is continuous and must be discretized before processing. After the signal had been amplified with a biopotential amplifier, it was sampled and quantized. The sampling theorem states that a continuous-time signal can be reconstructed from its samples if and only if the sampling frequency is greater than twice the signal bandwidth [21]. The minimum sampling frequency that allows the signal to be reconstructed is referred to as the Nyquist frequency. To avoid aliasing problems, it is necessary to band limit the signal to some frequency f_b before sampling. The Nyquist frequency then becomes $f_s = 2f_b$.

Quantization is the process of discretizing the magnitude of the signal. This can be accomplished by rounding or by truncation. Both methods introduce quantization error to the signal. This error can be modeled as additive, uniformly distributed white noise that is uncorrelated with the quantized signal [21]. It is also important to note that the signal-to-noise ratio of the quantization process increases by 6 dB for each added bit of digital coding. Therefore, it is advantageous to use as many bits as possible when digitally representing the signal to be processed. However, the number of bits used in the ocular prosthesis system will be determined by the hardware used in the prosthesis.

In acquiring the recorded EOG data, the signal was band-limited to 30 Hz by the biopotential amplifier's low-pass filter. The EOG signal itself is concentrated below 30 Hz, so no data was lost. The signal was sampled at 1 kHz, which is well above the Nyquist frequency of 60 Hz. The EOG signal was then quantized using 12 bits of digital coding, which was the maximum precision possible with the equipment used.

3.4 EOG Signal Processing Methods

The EOG signal must be processed after it has been amplified and digitized in order to de-noise the signal. This signal processing must be done without distorting the signal of interest if a true indication of the position of the natural eye is to be extracted from the EOG signal. Many signal processing methods have been used to de-noise the electrooculogram. Most of the examples of EOG processing in the literature come from eye movement detection applications in psychology. Work has also been done on processing the EOG signal to remove it from other biopotential signals such as the electroencephalogram.

In electroencephalogram research, the EOG is an undesirable signal that corrupts the signal of interest. Han and Park used independent component analysis to extract the EOG from the EEG signal [22]. Takano et al used a recursive least squares adaptive filter to perform the same task [23].

Other researchers have been interested in simply identifying the presence or absence of the EOG signal. Varri et al detected the beginning and end of saccades by calculating the standard deviation of the signal [24]. Varner et al used the short time Fourier transform to visually identify the presence of saccadic eye movements from the EOG signal [25]. Bonnet et al [26] and Buzenac et al [27] both used a Kalman filter to count the number of eye movements occurring in a sample of EOG data. Bonnet et al used hysteresis filters to complete the same task [26].

The traditional method for filtering the EOG signal is to apply a low-pass Butterworth filter [28] [29] [30]. This method works well for de-noising the signal, but it can distort the duration and velocity of the underlying signal. To address this prob-

lem, several more sophisticated filtering techniques are described in the literature. Hamer et al designed a unity gain, linear phase filter to de-noise the EOG signal [31]. Huebner et al [28] and Panse and Kulkarni [32] both used a Chebyshev type II filter to filter the signal. Engelken et al used a nonlinear smoothing filter [33], and Varri et al used a Kaiser-windowed FIR filter [24].

Adaptive filters have also been used. Bankman et al used an adaptive filter that resulted in no signal distortion [34]. However, this filter required a priori knowledge of the noise power, and assumed the signal to contain stationary, zero-mean additive noise. Zhu and Zhu used a similar adaptive filter, but their filter did not require a priori knowledge of the noise characteristics [30]. Park et al used autoregressive interpolation to replace sections of an EOG signal that contained ECG noise [29].

It is also important to note that few of the papers cited above mentioned how the EOG signal was amplified. A properly constructed biopotential amplifier will reject a significant amount of common mode noise in the EOG signal when the signal is amplified differentially. However, an improper amplifier would result in a noisy EOG signal that would require extensive processing to extract eye movement data from the signal. The lack of information about the amplification of the EOG signal in certain papers introduces some question as to the need for the more sophisticated methods used to process the EOG signal.

3.4.1 Processing Methods for Ocular Prosthesis System

The signal processing methods that are suitable for the ocular prosthesis project must fit the constraints imposed by the project. Specifically, the processing method chosen to de-noise the EOG signal should result in little distortion of the EOG signal

and must be implemented in the limited space and processing power available with the prosthesis, as mentioned in Section 1.2. Also, the signal processing must be done in real time.

The adaptive filter used by Bankman was not considered for de-noising the EOG in the ocular prosthesis application because the characteristics of the noise cannot be known a priori. Both this filtering technique and the techniques used by Zhu and Zhu and Engleken et al were too computationally intensive to be useful in the ocular prosthesis application. The technique of Park et al was not considered because it cannot be done in real time.

The methods that best suit the design constraints of the project are simple FIR and IIR filters. Therefore, an IIR Chebyshev type II filter [28] [32] and a Kaiser-windowed FIR filter [24] were evaluated for filtering the EOG signal for the ocular prosthesis project. These methods were compared to the IIR Butterworth filter, which is the usual filter used to de-noise the EOG, as mentioned above.

3.5 EOG Signal Processing Experiment

A signal processing experiment was conducted to evaluate which of the methods suggested in Section 3.4.1 (Butterworth, Chebyshev II, and Kaiser-windowed FIR filters) would be most appropriate for de-noising the EOG signal in the ocular prosthesis system. These methods were tested using a model saccade corrupted with real EOG noise, following the methods of Bankman and Thakor [34]. The use of the model saccade allowed the exact characteristics of the test signal to be known, which permitted quantitative analysis of the behaviour of each filter. An actual EOG

signal was not used to evaluate the signal processing methods because the characteristics of the noise present in the recorded signal cannot be known a priori. This precluded a quantitative analysis of the performance of each filter on an actual EOG recording. Once the best filtering method had been identified, it was used to process actual EOG data in order to verify visually that the chosen method was capable of de-noising the recorded EOG data.

3.5.1 Methods

The typical method of de-noising the EOG signal is to use a low-pass Butterworth filter. This method was compared with low-pass Chebyshev type-II and Kaiser-windowed FIR filters. The design criteria for all the filters were a low-pass cut-off frequency of 10 Hz, a transition region width of 10 Hz, a maximum attenuation in the pass band of 3 dB, and a minimum attenuation in the stop band of 40 dB.

The infinite impulse response filters (Butterworth and Chebyshev II) were implemented digitally in MatLab using the bilinear transformation. Butterworth filters are maximally flat in the passband and have a relatively wide transition region between the passband and stopband. Chebyshev II filters are flat in the passband and have equiripple in the stopband. The Chebyshev II filter has a narrower transition region than the Butterworth filter. The MatLab functions “buttord” and “cheby2ord” were used to determine the optimum filter order for the Butterworth and Chebyshev II filters, respectively, given the design criteria. The functions “butter” and “cheby2” were used to construct the filters. Figures 3.2 and 3.3 show the magnitude and phase response of the Butterworth and Chebyshev II filters, respectively.

The finite impulse response filter was implemented using a Kaiser window. FIR

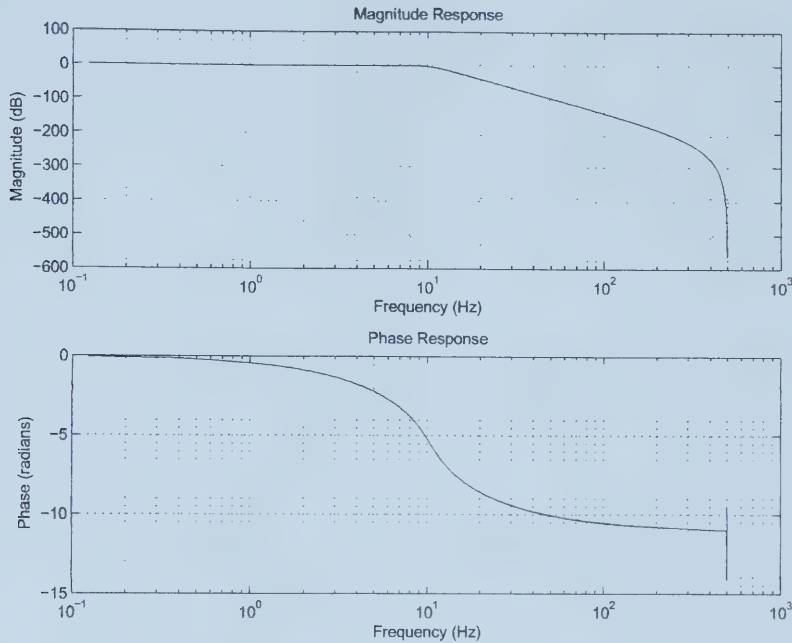


Figure 3.2: Magnitude and Phase Response of 7th-order Butterworth Filter

filters can be realized efficiently in hardware, but can result in a much higher filter order than the equivalent IIR filter. The MatLab function “`fir1`” constructs a filter of type I or type II, ensuring that the magnitude response of the filter at 0 Hz is non-zero, as is appropriate for a low-pass FIR filter. The Kaiser window was used because it provides the best trade-off between mainlobe width and sidelobe area. The length of the Kaiser window was determined using the MatLab function “`kaiserord`”, which determines the optimal window length for a given set of design criteria. Figure 3.4 shows the magnitude and phase response of the Kaiser-windowed FIR filter.

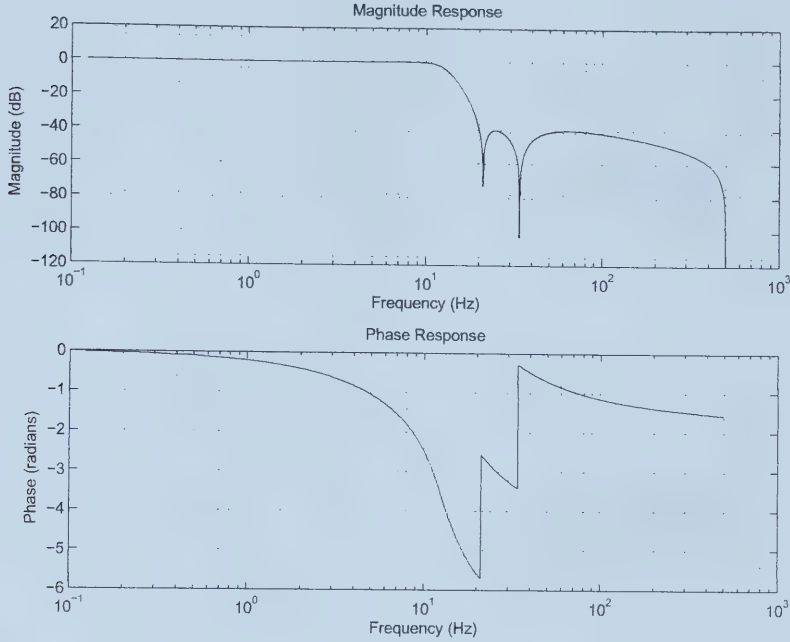


Figure 3.3: Magnitude and Phase Response of 5th-order Chebyshev II Filter

3.5.2 Generation of Model Saccade

A 10 degree model saccadic eye movement $s[k]$ was constructed as shown in Equation 3.1 and then contaminated with EOG noise [34]. The model saccade was based on a cosine wave, and is shown in Figure 3.5. The noise was obtained by recording the EOG while the subject's eyes were stationary using the methods outlined in Section 3.2. The EOG noise was then shifted in magnitude so that the signal was centred about 0 V. This data was then used to represent the contaminating noise present in the EOG recorded during eye movements. The extracted EOG noise $n[k]$ was added to the model saccade $s[k]$ to create the simulated noisy EOG waveform $x[k]$, shown

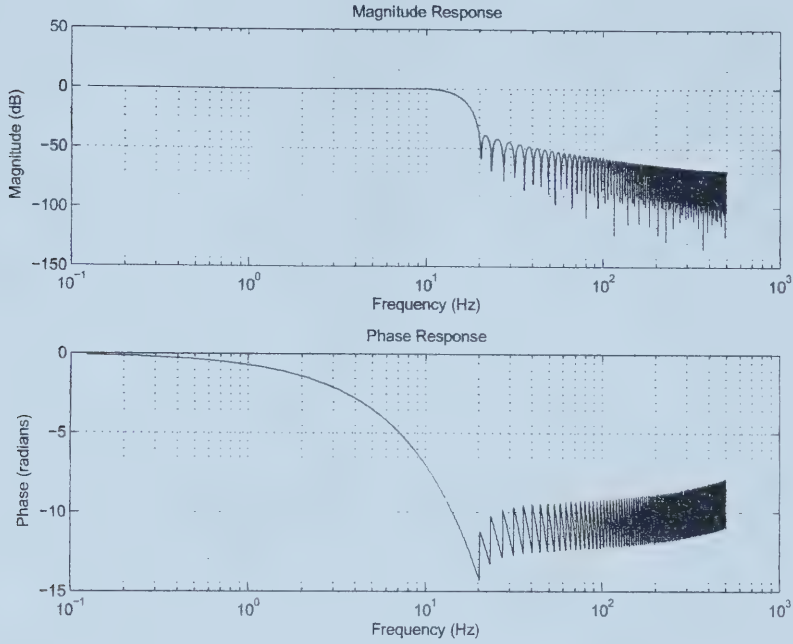


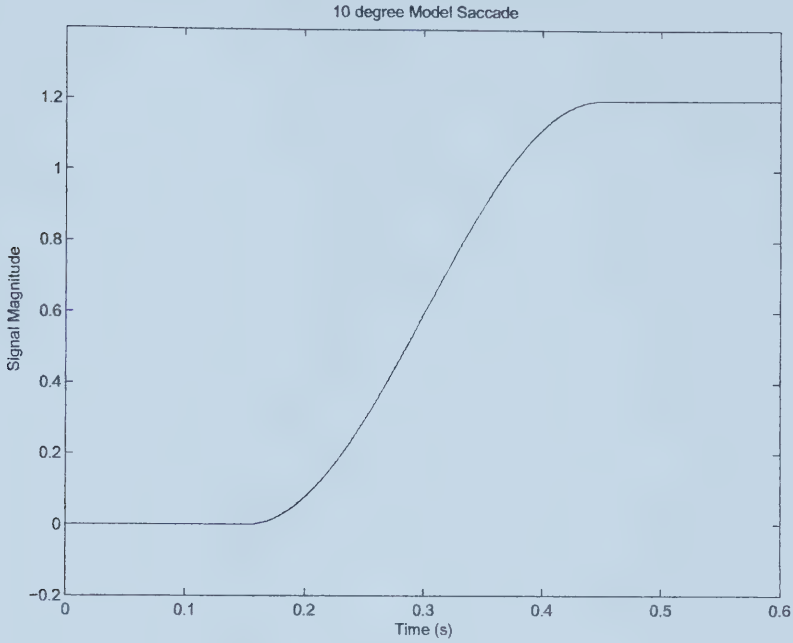
Figure 3.4: Magnitude and Phase Response of Kaiser-Windowed FIR Filter

in Figure 3.6.

$$s[n] = \begin{cases} 0 & 0.000s \leq n \leq 0.147s \\ 0.6 \cos \frac{2\pi n}{0.4} + 0.6 & 0.150s \leq n \leq 0.447s \\ 1.2 & 0.450s \leq n \leq 0.600s \end{cases} \quad (3.1)$$

3.5.3 Results of Processing Model Saccade

The Butterworth, Chebyshev II, and Kaiser-windowed FIR filters were evaluated using the simulated noisy EOG waveform described above. The filter order, eye

Figure 3.5: 10 degree Model Saccade $s[n]$

movement duration, and maximum lag of the filtered signal in each case were compared to that of the original model saccade in order to evaluate the performance of each filter. The eye movement duration was measured as the rise time of the saccadic eye movement. The lag was measured as the time difference between the midpoint (50% rise point) of the original saccade and the midpoint of the filtered saccade.

It can be seen from Table 3.1 that the FIR filter required a much higher order than the IIR filters to satisfy the design criteria. It also resulted in the lowest signal to noise ratio and the highest lag of all the filters tested. The IIR filters resulted in much better performance, with low filter orders and higher signal to noise ratios than

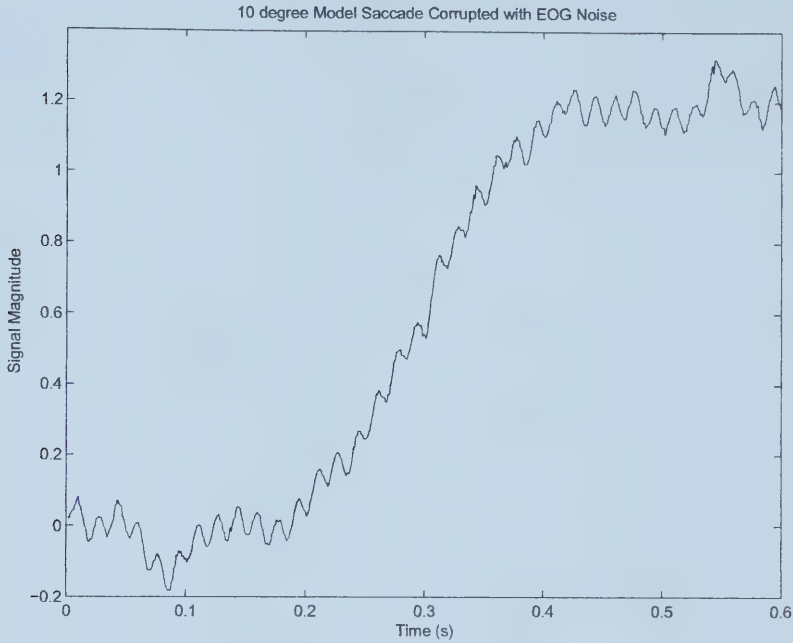


Figure 3.6: Model Saccade Corrupted with EOG Noise

the FIR filter. They also resulted in less lag than the FIR filter. This is important, since lag in the output signal would give an inaccurate indication of the position of the natural eye, resulting in the position of the ocular prosthesis lagging the position of the natural eye. The 5th-order Chebyshev II filter provided the best performance, and was used to filter actual EOG data, as discussed in the next section. Figure 3.7 shows the original model saccade as well as the output of the three filters. It was also found by inspection that all three filters introduced negligible distortion of the EOG signal duration or velocity, as can be seen in Figure 3.7.

	Original Saccade	Butterworth	Chebyshev Type II	Kaiser
Filter Order	n/a	7	5	223
SNR (dB)	23.573	26.155	26.836	24.786
Lag (s)	n/a	0.068	0.039	0.091

Table 3.1: Comparison of Filtering Methods

3.5.4 Results of Processing EOG Data

The 5th-order Chebyshev type II filter was used to de-noise actual EOG eye movement data. This data was recorded using the techniques described in Section 3.2, but under eye movement conditions. The subject was asked to look back and forth four times between targets placed 1 m in front of the subject, stopping to fixate for about 1 second on the target after each eye movement. Three eye movement trials were conducted at each of four different target spacings (2 degrees, 5 degrees, 10 degrees, and 30 degrees), for a total of 12 eye movement trials.

Figure 3.8 shows a section of an eye movement trials with 10 degrees target spacing, with the original signal on the top plot and the filtered signal on the bottom plot. It can be seen from the figure that the filtered signal closely follows the contours of the original signal but contains less noise.

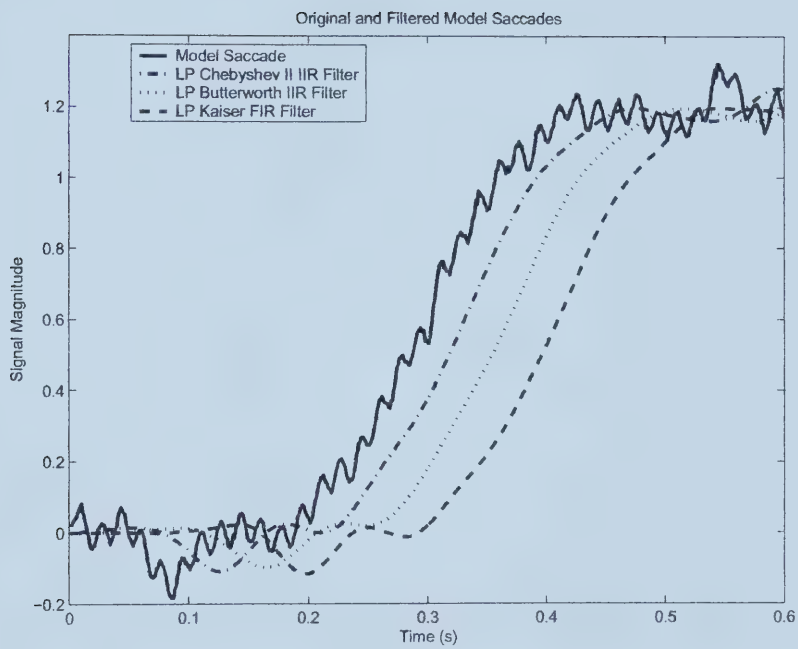


Figure 3.7: Original and Filtered Model Saccades

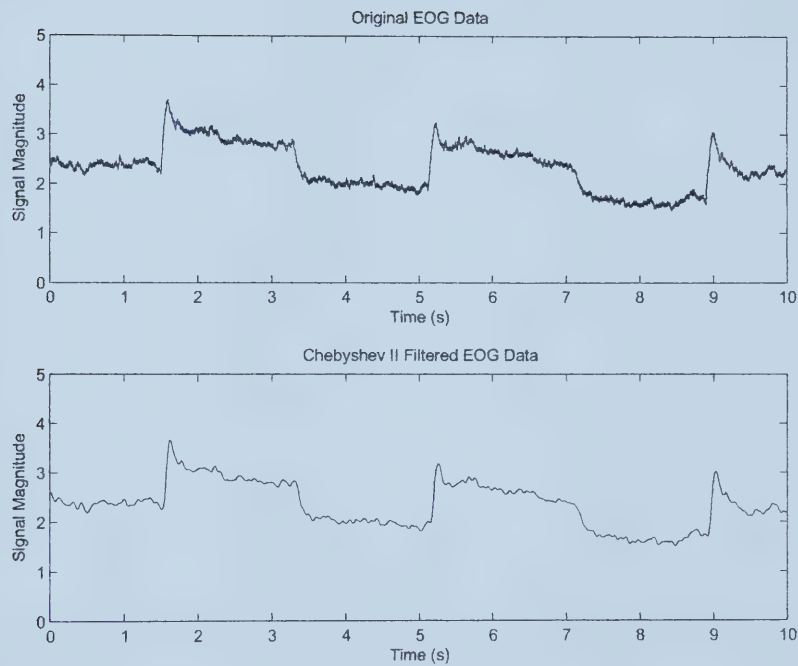


Figure 3.8: Original and Chebyshev II Filtered EOG Data

Chapter 4

Modeling the Source of the Electrooculogram

A numerical model of the source of the electrooculogram was constructed in order to determine the optimum placement of the implanted electrodes in the clinical ocular prosthesis system. As mentioned in Chapter 1, the electrooculogram is the potential sensed at the skin surface due to the ocular standing potential.

The development of the model of the ocular standing potential is described in Section 4.1. Section 4.2 describes how finite element analysis was used to predict the surface potentials that would be generated by the proposed ocular standing potential model. Sections 4.3 and 4.4 discuss how the model was verified by comparing the predicted surface potentials with actual surface potentials that were recorded experimentally.

4.1 Development of Ocular Standing Potential Model

The medical literature contains anatomical and physiological models of the mechanisms that give rise to the ocular standing potential, some of which are presented below. Several numerical models of the ocular standing potential are also discussed, and the model of the standing potential that was used in this thesis is presented.

4.1.1 Anatomical Models of the Ocular Standing Potential

Several anatomical models of the ocular standing potential have been proposed. It was discovered in 1865 that the retina is the anatomical source of the standing potential [35]. The retina consists of an electrically charged membrane, with the inner layers (towards the middle of the eye) being more positive than the outer layers (towards the edge of the eye). Without a change in light stimulation, the standing potential is constant, and has a value that ranges between 2 and 40 mV, depending on the individual [20].

The pigment epithelium layer of the retina is thought to be the most important source of the ocular standing potential [36] [37] [38] [39]. The photoreceptors of the neural layer of the retina likely contribute to the standing potential as well, but only when the retina is adapting to a changed light intensity (light or dark adaptation). The pigment epithelium theory is supported by several experiments. Babel et al noted that destroying the pigment epithelium significantly reduces the standing potential, while destroying the photoreceptors has no effect on the standing potential [20]. It has also been observed that the neural layer of the retina is dependent upon the central retinal artery for its circulation, whereas the pigment epithelium layer is dependent upon circulation through the choroidal layer of the eye. When the central retinal artery is occluded, the light-adapted ocular standing potential is reduced but the fast response of the standing potential (changes in the ocular standing potential due to eye movements) is unaffected [35]. This indicates that the portion of the standing potential not due to light adaptation originates in the layer of the retina not supplied by the central retinal artery, the pigment epithelium.

The pigment epithelium covers the posterior portion of the eyeball, and extends anteriorly to include the iris [40]. However, the pigment epithelium does not cover the optic disc, where the optic nerve connects to the eyeball. It is unclear from the ophthalmology literature whether the anterior portion of the pigment epithelium contributes to the ocular standing potential or not.

The electrooculogram is the potential sensed at the skin surface that is due to the ocular standing potential, as mentioned previously. The EOG varies as a function of eye position. This effect can be visualized by imagining the ocular standing potential to be an electric dipole oriented along the posterior-anterior axis of the eye, as shown in Figure 4.1.

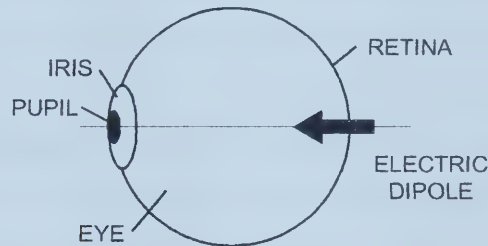


Figure 4.1: Ocular Standing Potential Visualized as a Dipole

The position of the eye can be determined by placing an electrode at each canthus of the eye and finding the difference between the recorded signals, with respect to some reference value. If the dipole is pointing directly forward the difference will be zero. If the eye rotates off centre the dipole will point more to one side than the other, and the difference between the electrode signals will not be zero.

For the purposes of modeling the ocular standing potential, the eye will be as-

sumed to be stationary. The ocular standing potential model is expected to be valid when the eye is moving as well as when the eye is stationary. However, there is currently not enough computing power available for this research to allow the standing potential model to be tested under eye movement conditions.

4.1.2 Numerical Models of the Ocular Standing Potential

The electric potential generated within the eye can be thought of as being distributed in a volume conductor (the eye). The electric field surrounding the eye is the sum of the fields due to all potential sources and sinks within the eye. The strength of the field surrounding the eye is dependent on the electric impedance of the tissues of the eye and its surrounding structures. For the purposes of the ocular prosthesis project, a model of the ocular standing potential was developed by hypothesizing the position, orientation, and strength of the potential sources and sinks that comprise the standing potential. The model was then verified by comparing measured surface potentials with predicted surface potentials generated using finite difference modeling. Section 4.1.3 discusses the model verification in more detail.

Several examples of ocular standing potential models can be found in the literature. Thickbroom and Mastaglia modeled the source of a pre-saccadic potential as a single dipole source located in the vicinity of the eye [41]. The pre-saccadic potential is a “spike” potential that occurs 10 to 40 ms before the onset of a saccadic eye movement. The head was modeled as a homogeneous spherical volume conductor, and electromagnetic field theory was used to calculate surface potentials due to the single dipole source. An iterative procedure that consisted of modifying the location, orientation, and strength of the source dipole was used to minimize the sum of the

squared differences between the predicted and measured surface potentials.

The methods used by Thickbroom and Mastaglia did not take into account the effect the different tissue types of the head would have on the distribution of surface potentials. The homogeneous nature of the head model used would have introduced errors to the surface potential distribution found. Also, the possibility of a distributed source was not considered.

Doslak, Plonsey, and Thomas used electromagnetic field theory to numerically model the ocular standing potential [42]. The eye was modeled as a heterogeneous volume conductor consisting of discrete homogeneous regions, where each region corresponded to an anatomical structure in the eye. The retina was treated as the source of the standing potential, and was modeled as a hemispherical dipole sheet of uniform strength. The authors assumed the eye to be spherical and axially symmetrical, which allowed the field calculations to be made for a two-dimensional projection of the eye into the transverse plane, instead of the three-dimensional case (see Figure 4.2). Approximately 1050 non-uniformly spaced nodes were distributed throughout the two-dimensional model of the eye. The spacing of the nodes was chosen empirically to provide the best compromise between accuracy, fast convergence, and the available memory (the calculations were performed in 1981 on a DEC PDP 1145 computer, which is now obsolete). The difference equation was solved at each node, accounting for the conductivities of the different regions of the eye and the variable nodal spacing.

The numerical model of the standing potential source was verified by comparing the numerical solution at each node with the analytic solution, for a simplified head model (three concentric homogeneous spheres). The conductivity values for this

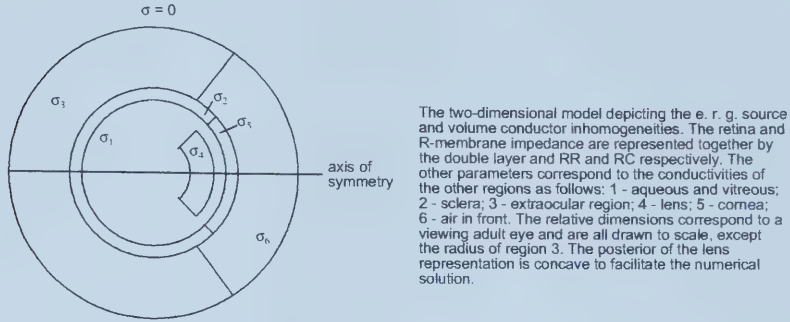


Figure 4.2: Heterogeneous Volume Conductor Model of the Eye [42]

experiment were obtained from the literature. The more accurate head model was evaluated by calculating the potential at various points along the anterior-posterior axis of the eye, and comparing the calculated potentials to values obtained from the literature.

The results of the work by Doslak et al may be more accurate than that of Thickbroom and Mastaglia, since a more realistic model of the eye was used. However, structures outside the eye were not included in the model. Also, the assumption that the eye is axially symmetric is an oversimplification, as the optic disc region of the eye lies to one side of the anterior-posterior axis of the eye. Compromises were also made in the verification of the model, which may have been due in part to the limited computing power available to the authors.

4.1.3 Proposed Model of the Ocular Standing Potential

The proposed model of the ocular standing potential addressed the shortcomings of the models used by Thickbroom and Mastaglia and Doslak et al. This model

used an anatomically accurate model of the head and investigated the possibility of a distributed source. The model was verified by comparison with experimentally recorded data.

The ocular standing potential was modeled as a non-uniform quasi-hemispherical dipole sheet. The thickness of the sheet was 1 mm, and its location corresponded to the location of the pigment epithelium layer of the retina, which is believed to be the major anatomical source of the standing potential. In reality, the thickness of the pigment epithelium is on the order of 1 μm ; however, the maximum resolution of the model in this case is 1 mm. To further increase the resolution of the model would result in a prohibitively time-consuming solution, given the facilities available for this research at the present time. Each dipole in the dipole layer pointed radially towards the centre of the eye. This configuration is supported by the radial orientation of the cells of the retina [43].

The portion of the dipole sheet corresponding to the optic disc did not contain dipole sources, as the pigment epithelium is not present in this area of the eye [40]. The diameter of the optic disc region was varied experimentally between 0 mm and 5 mm. The extent to which the dipole sheet extended anteriorly toward the iris was varied to determine the optimum model. Also, the magnitude of the dipoles was varied across the dipole sheet to assess whether or not those areas of the pigment epithelium that receive more incident light contribute more strongly to the ocular standing potential.

For the purposes of constructing the ocular standing potential model, the eye was assumed to be a sphere of diameter 25 mm. A hemispherical shell $S(P(x, y, z))$ conforming to these dimensions was constructed where the dipole magnitude at each

point was 1, as given in Equation 4.1.

$$S(P(x, y, z)) = 1 \quad \forall P(x, y, z) \text{ such that } \sqrt{x^2 + y^2 + z^2} = \frac{25}{2} \quad (4.1)$$

The set of points comprising the shell $S(P(x, y, z))$ was assumed to be the set of sources for the dipoles in the dipole sheet. For each point in $P(x, y, z)$, an ideal sink point was also found. Figures 4.3 and 4.4 show the source and sink shells that contain the source point $P(x_0, y_0, z_0)$ and the ideal sink point $P(x_1, y_1, z_1)$, respectively, where the magnitude of $P(x_1, y_1, z_1)$ was -1. Since the resolution of the model was 1 mm, the length of all dipoles in the dipole sheet was assumed to be 1 mm. The direction of each dipole was from the source point towards the centre of the eye, as previously mentioned.

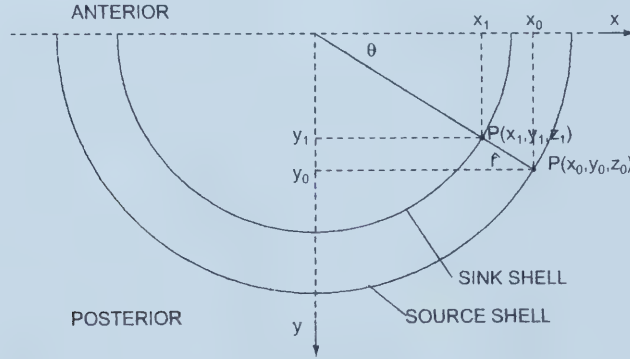
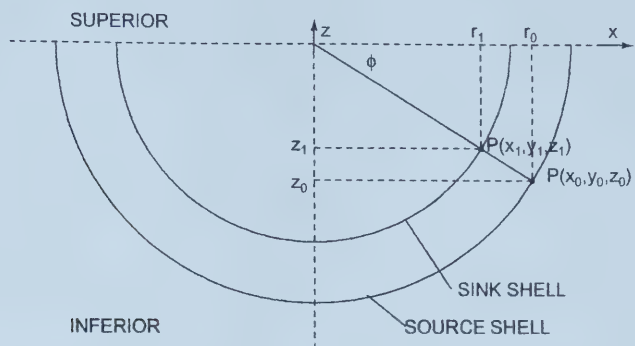


Figure 4.3: Construction of x - and y -components of Current Dipole

The resolution of the model does not allow all of the ideal sink points to be included in the model. Therefore, three sink points were created for each source point, corresponding to the x -, y -, and z -components of the dipole between the

Figure 4.4: Construction of z -component of Current Dipole

source and ideal sink points. These three sink points are $P(x_1, y_0, z_0)$, $P(x_0, y_1, z_0)$, and $P(x_0, y_0, z_1)$, where x_1 , y_1 , and z_1 are given in Equations 4.2 through 4.4,

$$x_1 = x_0 + c \quad (4.2)$$

$$y_1 = y_0 + b \quad (4.3)$$

$$z_1 = z_0 + a \quad (4.4)$$

where

$$\begin{aligned}
 a &= \sin \phi \\
 b &= \hat{r} \sin \theta \\
 c &= \hat{r} \cos \theta \\
 \text{and } \hat{r} &= \cos \phi \\
 \theta &= \tan^{-1} \left(\frac{y_0}{x_0} \right) \\
 \phi &= \sin^{-1} \left(\frac{z_0}{\sqrt{x_0^2 + y_0^2 + z_0^2}} \right)
 \end{aligned}$$

The quantities a , b , and c are shown in Figure 4.5. The magnitude at the sink points $P(x_1, y_0, z_0)$, $P(x_0, y_1, z_0)$, and $P(x_0, y_0, z_1)$ was c , b , and a , respectively, such that the vector sum of the component dipoles was the ideal dipole between the source point and the ideal sink point.

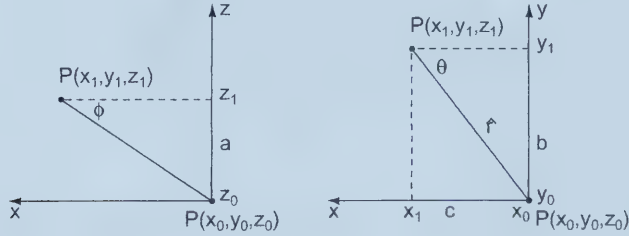


Figure 4.5: Construction of a , b , and c

The angular distance the dipole sheet extended forward along the retina was varied between 30 and 90 degrees with respect to the anterior-posterior axis of the eye, as shown in Figures 4.6 and 4.7. The dipole sheet is shown with the posterior pole of the sheet pointing towards the top of the page, due to the limitations of the software used to generate these figures. This set of variations to the dipole

sheet model was introduced to evaluate the degree to which incident light on the retina contributes to the ocular standing potential; if the models having 90 degrees of anterior extension were more accurate than those having 30 degrees of anterior extension, then it could be assumed that incident light contributes to the ocular standing potential.

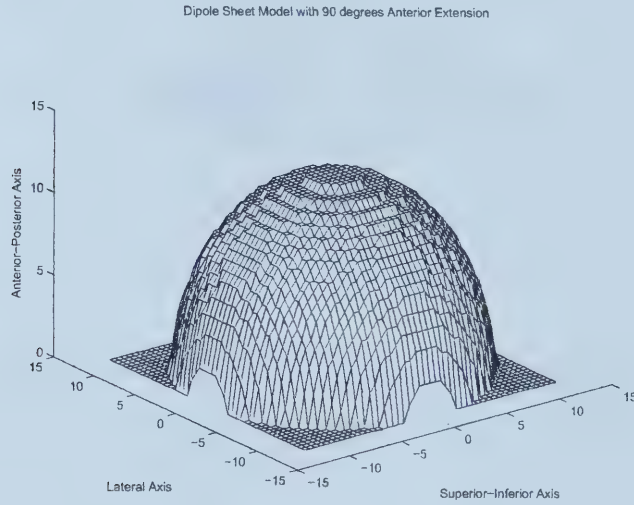


Figure 4.6: Dipole Sheet Model with 90 degrees Anterior Extension

The diameter of the optic disc region of the model was varied between 0 mm and 5 mm, since the exact diameter of the area missing the pigment epithelium was not clear from the ophthalmology literature. Figure 4.8 shows a plot of the dipole sheet with a 3 mm optic disc region removed. Again, the dipole sheet is shown with its posterior pole pointing towards the top of the page.

The distribution of dipole magnitude was also varied. Models having uniformly

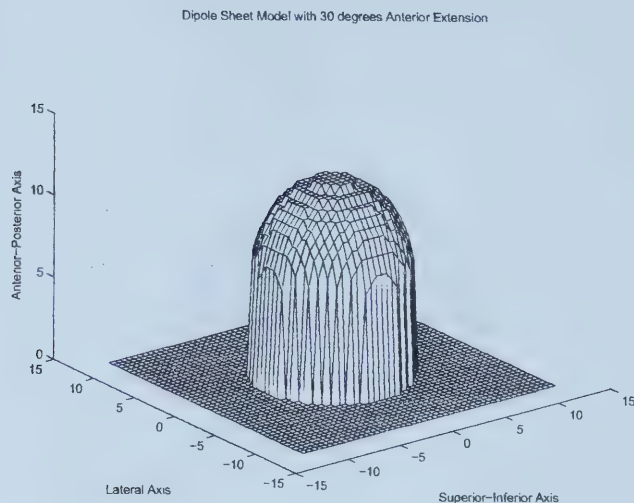


Figure 4.7: Dipole Sheet Model with 30 degrees Anterior Extension

distributed, linearly decreasing, quadratically decreasing, and normally distributed dipole magnitudes were evaluated. In each case, the maximum dipole magnitude was located at the posterior pole of the eye and decreased, if at all, as the dipole sheet extended forward. These distributions were intended to simulate possible distributions of incident light on the retina.

The uniform, quadratic, and normal distributions were used in order to evaluate the effect of incident light intensity on the ocular standing potential. On the level of a single cell of the pigment epithelium, it is not known whether or not the magnitude of the cell's response to incident light is a linear function of light intensity. Models with a non-linear distribution of dipole magnitude were used to account for this unknown.

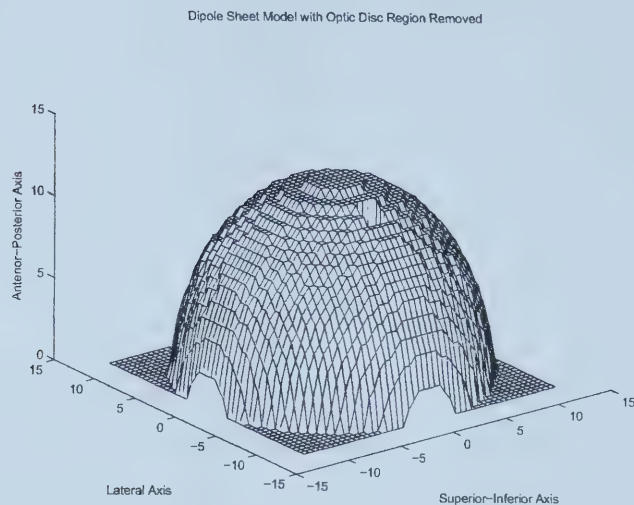


Figure 4.8: Dipole Sheet Model with Optic Disc Region Removed

The normally distributed model was assumed to have a standard deviation of 3.7, which is approximately one third the radius of the eye.

Equations 4.5 through 4.8 give the dipole magnitude as a function of x and z for the uniform, linear, quadratic, and normal distributions, respectively, where $r_{max} = 12.5$ mm is the radius of the eye, $\sigma = 3.7$, and the x - z plane is the coronal (frontal) plane of the head. Since the dipole sheets are quasi-hemispherical, all points in a given x - z plane will be of equal dipole magnitude. Therefore, y does not appear in these equations, where y lies along the anterior-posterior axis of the eye.

$$M_{uniform}(x, z) = 1 \quad (4.5)$$

$$M_{linear}(x, z) = \frac{r_{max} - \sqrt{x^2 + z^2}}{r_{max}} \quad (4.6)$$

$$M_{quadratic}(x, z) = \frac{r_{max}^2 - \sqrt{x^2 + z^2}}{r_{max}^2} \quad (4.7)$$

$$M_{normal}(x, z) = \frac{\frac{1}{\sqrt{2\pi}\sigma} e^{-\frac{(\sqrt{x^2+z^2})^2}{2\sigma^2}}}{r_{max}} \quad (4.8)$$

For each model, the plots in Figure 4.9 indicate the dipole strength at each point in a transverse slice through the middle of the eye, as a function of x . The x -axis represents the lateral axis of the eye, and the y -axis represents dipole strength.

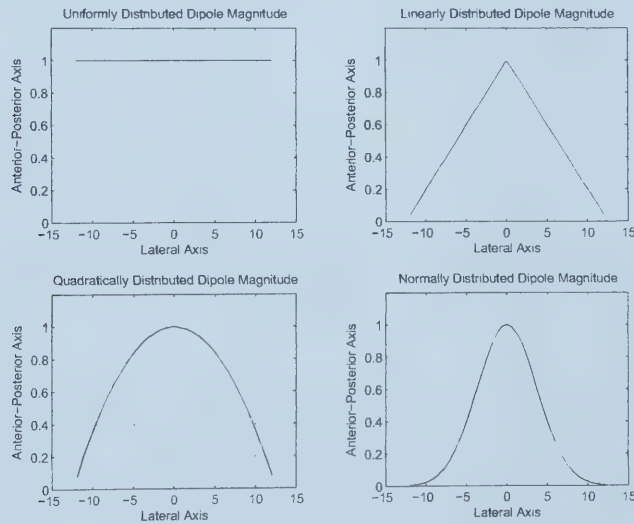


Figure 4.9: Distribution of Dipole Strength

A single dipole model of the ocular standing potential was also constructed, in

order to compare the distributed source models with a single source model. The single dipole was located at the posterior pole of the eye and pointed towards the centre of the eye. Its strength was 1000, which was experimentally determined to be the best value for comparison with the distributed models, during initial simulations, since this strength of single dipole resulted in surface potentials of a similar magnitude to those measured experimentally.

4.2 Generation of Predicted Surface Potentials

The surface potentials at ten locations around the natural eye were predicted for each proposed model of the ocular standing potential. The finite element method that was used to generate these predictions is discussed below, as is the development of the head model that was used. The numerical methods employed to solve for the surface potentials are also presented.

4.2.1 Finite element modeling

Finite element modeling involves modeling the volume conductor as an assemblage of equally sized elements, or voxels. In this case, the volume conductor of interest is the human head. The voxels in the model had a volume of 1 cubic mm. The development of this model is discussed in greater detail in Section 4.2.2. The conductivity of each voxel was determined by the conductivity of the tissue type to which that voxel corresponded.

The voxels can be thought of as comprising a linear resistive network, where the centre of each voxel is represented by a node in the network. The conductivity of the

connection between two adjacent nodes in the network is the series combination of the conductivities of the adjacent nodes, as given in Equation 4.9. The relationship between the voxel assemblage and the resistive network is shown in Figure 4.10.

$$\sigma_{\text{effective}} = \left[\frac{1}{\sigma_0/2} + \frac{1}{\sigma_1/2} \right]^{-1} = \frac{2\sigma_0\sigma_1}{\sigma_0 + \sigma_1} \quad (4.9)$$

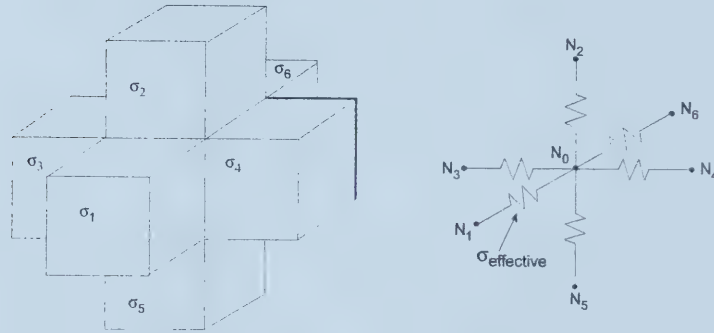


Figure 4.10: Relationship Between Voxel Assemblage and Resistive Network

If a current source is located at a specific node, the potential at any node in the network can be calculated using Kirchoff's current law, which states that the sum of currents into a node is zero, unless that node is a current source (or sink) (see Equation 4.10) [44].

$$\frac{I_0}{h} = \sum_{i=1}^6 \sigma_{\text{effective},i} (V_i - V_0) \quad (4.10)$$

where V_0 is the potential of the node of interest

V_i is the potential at each adjacent node

I_0 is the total current into the node of interest

h is the side length of the voxel (i.e. 1 mm)

σ is the conductivity between the node of interest and node i

Equation 4.11 states Kirchoff's current law in matrix form, where $\mathbf{A}$ is the conductivity matrix, $\mathbf{v}$ is the vector of potentials at each node, and $\mathbf{i}$ is the vector containing the sum of currents entering each node. The current vector $\mathbf{i}$ is sparse, and the conductivity matrix is symmetric and sparse [45].

$$\mathbf{A}\mathbf{v} = \mathbf{i} \quad (4.11)$$

The matrix equation can be solved using a method such as the conjugate gradient method [45]. The conjugate gradient method is a numerical method for solving an $N \times N$ linear system of equations, given in Equation 4.12.

$$\mathbf{A}\mathbf{x} = \mathbf{b} \quad (4.12)$$

If the matrix $\mathbf{A}$ is symmetric and positive definite, then the system in Equation

4.12 can be solved by minimizing

$$f(\mathbf{x}) = \frac{1}{2} \mathbf{x} \mathbf{A} \mathbf{x} - \mathbf{b} \mathbf{x} \quad (4.13)$$

which is minimized when its gradient, given in Equation 4.14, is zero; in this case, Equation 4.14 is equivalent to Equation 4.12.

$$\nabla f = \mathbf{A} \mathbf{x} - \mathbf{b} \quad (4.14)$$

The conjugate gradient method minimizes the function $f(\mathbf{x})$ by generating search directions $\{\mathbf{p}_k\}$ and values $\{\alpha_k\}$, where the values $\{\alpha_k\}$ are found such that they minimize $f(\mathbf{x}_k + \alpha_k \mathbf{p}_k)$. The next value of $\mathbf{x}_k$ is calculated using Equation 4.15, making $\mathbf{x}_{k+1}$ a better minimizer of $f(\mathbf{x})$ than $\mathbf{x}_k$.

$$\mathbf{x}_{k+1} = \mathbf{x}_k + \alpha_k \mathbf{p}_k \quad (4.15)$$

The value $\mathbf{x}_{k+1}$ is chosen so that it also minimizes $f(\mathbf{x})$ over the whole vector space of directions $\{\mathbf{p}_1, \mathbf{p}_2, \dots, \mathbf{p}_k\}$ already chosen.

It is important that the conductivity matrix is well conditioned when using the conjugate gradient method, as this will result in faster convergence than a poorly conditioned $\mathbf{A}$ matrix. If the conductivity matrix is not well conditioned, it can be improved by using a method such as polynomial pre-conditioning [46]. In this method, an ideal matrix $\mathbf{P}$ is used to pre-multiply the system in Equation 4.12, as shown in Equation 4.16.

$$\mathbf{PAv} = \mathbf{Pi} \quad (4.16)$$

$\mathbf{P}$ is chosen such that

$$\mathbf{PA} = \mathbf{I} \quad (4.17)$$

Obtaining the ideal matrix $\mathbf{P}$ would require inverting the $\mathbf{A}$ matrix, which is impractical for source localization problems due to the very large size of the $\mathbf{A}$ matrix. Instead, a matrix $\hat{\mathbf{P}}$ is constructed as given in Equation 4.18 such that $\hat{\mathbf{P}} \approx \mathbf{P}$.

$$\hat{\mathbf{P}} = \beta_0 \mathbf{I} + \beta_1 \mathbf{A} + \beta_2 \mathbf{A}^2 + \dots \approx \mathbf{P} \quad (4.18)$$

It was found by trial and error that pre-conditioning using 7th order polynomials was effective in producing a well conditioned conductivity matrix for the EOG source localization problem.

A dipole source can be modeled as two sources of equal and opposite magnitude at different nodes (the source and the sink). The magnitude and orientation of the dipole is determined by the relative locations of the sink and source. Determining the potential at the nodes corresponding to locations on the surface of the volume conductor, due to a dipole source, is known as the forward problem in source localization. Finding the location and orientation of a current source by measuring the surface potentials is the inverse problem.

The inverse solution of the source localization problem is used clinically to de-

termine the origin of electrochemical activity in the body. For example, the source of an epileptic seizure can be determined by measuring the potentials at the surface of the head due to the epileptic activity, and then solving the inverse problem to localize the source of the seizure [11].

The inverse problem can be solved by comparing measured potentials on the surface of the volume conductor with calculated potentials. The lead field matrix, given in Equation 4.19, is the matrix of potentials induced at each site of interest on the surface of the volume conductor by a current dipole at each voxel and orientation in the volume conductor. Each entry V_{ij} in the lead field matrix gives the potential at a specific location on the surface of the volume conductor, due to a current dipole at a specific location inside the volume. For a given electrode location, the potential at that location due to a current dipole at each voxel and orientation can be determined by extracting the information from the lead field matrix. For each current dipole considered (i.e. for a current dipole at each possible voxel and orientation), the least squares difference between the recorded potential and the calculated potential is found. The current dipole that results in the minimum least squares difference measure, across all the electrode locations, corresponds to the correct source voxel.

$$\mathbf{L} = \begin{bmatrix} \mathbf{V}_{1x} \\ \mathbf{V}_{1y} \\ \mathbf{V}_{1z} \\ \vdots \\ \mathbf{V}_{Nx} \\ \mathbf{V}_{Ny} \\ \mathbf{V}_{Nz} \end{bmatrix}, \quad \mathbf{V}_{ix} = \{V_{ix,1}, \dots, V_{ix,M}\} \quad (4.19)$$

where N is number of voxels

M is number of sites on surface of volume conductor

However, computation of the lead field matrix is in general too time consuming. For example, a system with ten electrodes and a volume conductor consisting of 4 million voxels would require the forward problem to be solved 12 million times to construct the lead field matrix. To solve this problem, the principle of reciprocity is applied to the linear resistive network. A current source is applied to each electrode location on the surface of the volume conductor and a current sink is applied to the reference electrode. The forward problem is solved to generate the resulting voltage at every voxel. By taking the potential differences across every pair of voxels in each of the three orthogonal directions, the column-wise lead field matrix is generated, as shown in Equation 4.20. This method requires that the forward problem be solved only 10 times, for the example given above.

Solution of the inverse problem using the principle of reciprocity assumes that

the source consists of a single dipole. Therefore, this method was not used to localize the ocular standing potential since the ophthalmology literature suggests that this potential is a distributed source. Instead, a series of models of the ocular standing potential were hypothesized. The surface potentials resulting from each model were compared to recorded surface potentials. The model resulting in the lowest sum of root mean squared errors between the predicted and recorded surface potentials then represented the most accurate model of the ocular standing potential.

Inverse methods for generating a distributed source are available, but are presently too computationally intensive to be useful for this application.

$$\mathbf{L} = \{\mathbf{V}_{E1}, \dots, \mathbf{V}_{EM}\}, \quad \mathbf{V}_{Ei} = \begin{bmatrix} V_{1x} \\ V_{1y} \\ V_{1z} \\ \vdots \\ V_{Mx} \\ V_{My} \\ V_{Mz} \end{bmatrix} \quad (4.20)$$

4.2.2 Construction of Head Model

The standard head model used in biopotential source localization of the head is a concentric spherical shell model [11] [46]. However, this method is known to be inaccurate, and can result in source localization errors. Therefore, it is desirable to use an anatomically accurate head model [7] [8] [47] [48].

The finite element model of the head was constructed from the anatomical data

contained in the Body Explorer 2.0 CD atlas. This resource consists of colour photographs of 1 mm transverse slices of a human cadaver. The slices pertaining to the head were segmented into different tissue types using colour segmentation of the photographs as well as edge tracing. Some manual correction of the images was also necessary in places. Figures 4.11 and 4.12 show the original photographic data and the segmented image for a transverse slice through the eyes, respectively. It should be noted that there are some discrepancies between the photographic data and the segmented data. This is due to problems with the segmentation algorithm that was used to generate the segmented data. However, this head model is significantly more accurate than the concentric spherical shell model, and as such is an improvement over the standard modeling technique.

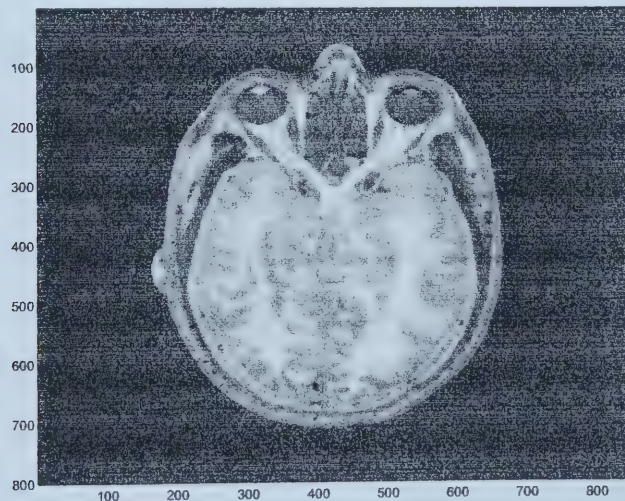


Figure 4.11: Original Photographic Data

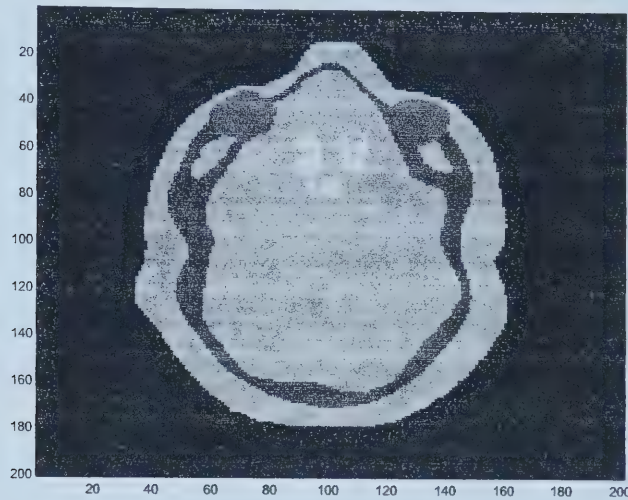


Figure 4.12: Segmented Data

The tissue types used were soft tissue (including skin, fat, muscle, and connective tissue), brain matter (including white and grey matter), bone, eye, and air-filled sinuses (frontal, sphenoid, ethmoid, and maxillary sinuses and nasal cavity). The sinuses were included in the model because of their proximity to the ocular standing potential. Cuffin [49] and van Broek [10] both found that inaccurate modeling of areas of low conductivity (such as the sinuses) that are located near the biopotential source can result in source localization errors. Fender found this to be true for sources located near inaccurately modeled sinuses, in particular [48]. For this reason, care was taken to ensure that the air-filled sinuses in the segmented data were modeled accurately. In this context, inaccurate modeling refers to anatomical structures being modeled by regions of a volume and/or position that do not correctly represent the

structures in question.

The conductivity values for each tissue type were determined from the literature, and are given in Table 4.1 [50] [51] [52].

Tissue	Conductivity ($1/\Omega m$)
External Air	0.000
Internal Air	0.002
Bone	0.013
Soft Tissue	0.173
Brain	0.303
Eye	0.500

Table 4.1: Conductivity Values

Each 1 mm slice was broken into 1 cubic mm voxels, as this was the resolution of the finite element model. The resulting head model consisted of approximately 4 million voxels. The models of the ocular standing potential were each incorporated into the head model such that the standing potential model corresponded to the location of the retina of the right eye. The result was a series of head models, each including a different model of the ocular standing potential. The reference element was located at the chin in each model.

4.2.3 Solution Method

The head models described above were used to calculate the predicted potentials at ten locations on the surface of the head. These locations were in the vicinity of the eye, and corresponded to the locations of the electrodes used to record the actual surface potentials. The surface potentials were recorded at each of these electrode locations during an attempted steady gaze. That is, the eye was assumed to be

stationary. The locations of the electrodes and the data collection procedures are described in Section 3.2. For each head model, the root mean squared error between the predicted and measured surface potentials was calculated. The sum of the root mean squared errors for each head model was then found, and the head model giving rise to the minimum sum was considered to contain the most accurate model of the ocular standing potential. This method was used in place of the solution of the inverse problem, which generates a single dipole source, not the distributed source suggested by the anatomical models of the standing potential.

4.2.4 Simulation Methods

The data files necessary for generating the predicted surface potentials for each head model were generated using MatLab on a Sun Microsystems Ultra 30 running Unix with 2 GB of RAM. The files were then transferred to a second machine for use in the solution of the forward problem. These simulations were performed on an HP RX4610 running Linux on four Intel Itanium processors with 8 GB of RAM. It was not possible to generate the data files on the HP machine because MatLab is not currently available for the Itanium machine.

The forward problem required approximately 105 minutes to converge to a solution on the Itanium machine, running on one processor, and required 380 MB of RAM. Polynomial pre-conditioning using 7th order polynomials was used in order to improve the condition of the conductivity matrix before beginning the conjugate gradient algorithm. Convergence occurred in under 220 iterations in all cases, and resulted in a solution residual of less than 1.3E-9.

4.2.5 Simulation Results

Table 4.2 lists the head models for which the forward solution was found. The models are identified by the ocular standing potential model that they contain. The results of the simulations, the surface potentials at the ten electrode sites around the natural eye, are given in Tables 4.3 through 4.7. Table 4.8 contains metrics that were used to evaluate the results of these simulations. The standard deviation of the surface potential values was used to evaluate the range of potentials generated by each head model. This was a useful tool for determining whether or not a model was providing a sufficiently accurate contour of surface potentials. The mean value of the predicted potentials for each model was used to assess the overall accuracy of each model. These metrics were simply used to understand the nature of the results for each model. The sum of the mean squared errors between the simulated and actual surface potentials was used to determine the best model.

Anterior Extension	Optic Disc Diameter (mm)			
30	0	1	3	5
60	0	1	3	5
90	0	1	3	5

Table 4.2: Ocular Standing Potential Models (Anterior Extension vs. Optic Disc Diameter)

4.3 Collection of Recorded Surface Potentials

The actual surface potentials were recorded using the data collection techniques described in Section 3.2. The electrodes were placed at ten sites in the vicinity of

		Solution at Each Electrode Location									
AE	OD	1	2	3	4	5	6	7	8	9	10
30	0 mm	0.166	0.185	0.190	0.201	0.173	0.161	0.177	0.184	0.160	0.146
60	0 mm	0.998	1.128	1.167	1.241	1.049	0.971	1.081	1.132	0.964	0.881
90	0 mm	1.874	2.127	2.201	2.356	1.981	1.826	2.024	2.122	1.811	1.650
30	1 mm	0.164	0.182	0.187	0.198	0.171	0.159	0.175	0.182	0.158	0.145
60	1 mm	0.996	1.126	1.165	1.238	1.047	0.969	1.079	1.130	0.962	0.879
90	1 mm	1.872	2.125	2.198	2.353	1.979	1.825	2.021	2.120	1.809	1.648
30	3 mm	0.145	0.162	0.167	0.177	0.153	0.142	0.156	0.162	0.141	0.129
60	3 mm	0.978	1.106	1.144	1.217	1.029	0.951	1.059	1.110	0.945	0.863
90	3 mm	1.854	2.105	2.178	2.332	1.960	1.807	2.002	2.100	1.791	1.633
30	5 mm	0.119	0.134	0.138	0.146	0.126	0.117	0.128	0.134	0.116	0.106
60	5 mm	0.936	1.060	1.097	1.169	0.986	0.911	1.015	1.064	0.906	0.827
90	5 mm	1.813	2.059	2.131	2.283	1.917	1.767	1.958	2.054	1.752	1.596

Table 4.3: Forward Solution for Uniformly Distributed Dipole Sheet Models (AE - Anterior Extension, OD - Diameter of Optic Disc)

the eye, as shown in Figure 4.13. Electrodes 1-6 were evenly spaced around the eye, and electrodes 7-10 were evenly spaced in an arc above and to the side of the eye, which is not shown in Figure 4.3. These electrode placements were used to record surface potentials in three dimensions from the vicinity of the eye, and were based on the idea of the international 10-20 system of electrode placement used in electroencephalogram recording [17].

The signals were recorded referentially with respect to the reference electrode, which was placed on the chin. Twenty trials of 10 seconds each were performed and recorded, with as much time between the trials as the subject felt was necessary.

AE	OD	Solution at Each Electrode Location									
		1	2	3	4	5	6	7	8	9	10
30	0 mm	0.104	0.116	0.119	0.126	0.109	0.101	0.111	0.115	0.100	0.092
60	0 mm	0.336	0.379	0.392	0.416	0.353	0.327	0.363	0.380	0.325	0.297
90	0 mm	0.377	0.426	0.440	0.467	0.397	0.367	0.408	0.426	0.364	0.333
30	1 mm	0.103	0.114	0.117	0.124	0.108	0.100	0.110	0.114	0.099	0.091
60	1 mm	0.336	0.379	0.392	0.416	0.353	0.327	0.363	0.380	0.325	0.297
90	1 mm	0.376	0.424	0.438	0.466	0.395	0.366	0.406	0.425	0.363	0.332
30	3 mm	0.091	0.102	0.105	0.111	0.096	0.089	0.098	0.102	0.088	0.081
60	3 mm	0.324	0.365	0.378	0.401	0.341	0.315	0.350	0.366	0.313	0.286
90	3 mm	0.365	0.412	0.426	0.452	0.384	0.355	0.394	0.413	0.353	0.322
30	5 mm	0.075	0.083	0.086	0.091	0.079	0.073	0.080	0.083	0.072	0.066
60	5 mm	0.300	0.339	0.351	0.374	0.316	0.292	0.325	0.340	0.291	0.265
90	5 mm	0.341	0.386	0.399	0.425	0.359	0.332	0.369	0.387	0.330	0.301

Table 4.4: Forward Solution for Linearly Decreasing Dipole Sheet Models (AE - Anterior Extension, OD - Diameter of Optic Disc)

4.3.1 Collected Data

The data from each trial was passed through a 5th-order Chebyshev Type II low-pass filter with a cut-off frequency of 10 Hz. The attenuation of the filter in the passband was 3 dB, and the attenuation in the stopband was 40 dB. The Chebyshev II filter was chosen because it was found to be the most suitable filter for de-noising the EOG in the ocular prosthesis system, as described in Chapter 3.

Ten data sets were selected for use in verifying the proposed ocular standing potential models. The criteria used for selection of these data sets were a small amount of drift in the signals, few artifacts due to skin resistance and electrode movement.

Figure 4.14 shows an example of the filtered signals obtained from each of the ten sensing electrodes, for one of the ten data sets selected. The mean values of the

AE	OD	Solution at Each Electrode Location									
		1	2	3	4	5	6	7	8	9	10
30	0 mm	0.141	0.157	0.162	0.171	0.148	0.137	0.151	0.157	0.136	0.125
60	0 mm	0.533	0.601	0.622	0.660	0.560	0.519	0.576	0.603	0.515	0.471
90	0 mm	0.611	0.690	0.713	0.758	0.642	0.594	0.661	0.691	0.590	0.539
30	1 mm	0.140	0.156	0.160	0.169	0.146	0.136	0.149	0.155	0.135	0.124
60	1 mm	0.532	0.600	0.620	0.659	0.559	0.517	0.575	0.601	0.514	0.470
90	1 mm	0.610	0.688	0.711	0.756	0.641	0.593	0.659	0.689	0.589	0.538
30	3 mm	0.133	0.149	0.153	0.162	0.140	0.130	0.143	0.148	0.129	0.118
60	3 mm	0.526	0.593	0.613	0.651	0.552	0.511	0.568	0.594	0.508	0.464
90	3 mm	0.594	0.671	0.694	0.738	0.625	0.578	0.642	0.672	0.574	0.524
30	5 mm	0.120	0.133	0.137	0.144	0.125	0.116	0.128	0.132	0.115	0.106
60	5 mm	0.507	0.572	0.591	0.627	0.532	0.493	0.548	0.573	0.490	0.447
90	5 mm	0.561	0.634	0.656	0.699	0.591	0.546	0.607	0.636	0.543	0.495

Table 4.5: Forward Solution for Quadratically Decreasing Dipole Sheet Models (AE - Anterior Extension, OD - Diameter of Optic Disc)

ten electrode signals for each of the ten data sets are given in Table 4.9. These are the values which were used to verify the accuracy of each ocular standing potential model. The standard deviations and means of the ten signals are given for each data set in Table 4.10, for the purposes of comparison with the predicted surface potentials.

4.4 Analysis of Results

For each ocular standing potential model and each recorded data set, the sum of the root mean squared errors between the predicted surface potentials and the recorded potentials was calculated. The sums of the root mean squared errors for a given standing potential model were totalled across the ten recorded data sets to yield an accuracy score for each standing potential model. The accuracy scores ranged

AE	OD	Solution at Each Electrode Location									
		1	2	3	4	5	6	7	8	9	10
30	0 mm	0.090	0.100	0.103	0.109	0.094	0.087	0.096	0.099	0.087	0.079
60	0 mm	0.187	0.209	0.216	0.229	0.196	0.181	0.201	0.210	0.180	0.165
90	0 mm	0.203	0.229	0.236	0.251	0.214	0.198	0.219	0.229	0.197	0.180
30	1 mm	0.089	0.099	0.101	0.107	0.093	0.086	0.095	0.098	0.086	0.078
60	1 mm	0.335	0.378	0.390	0.414	0.352	0.326	0.362	0.378	0.324	0.296
90	1 mm	0.202	0.228	0.235	0.249	0.213	0.197	0.218	0.228	0.196	0.179
30	3 mm	0.079	0.088	0.091	0.096	0.083	0.077	0.085	0.088	0.076	0.070
60	3 mm	0.176	0.198	0.204	0.217	0.185	0.171	0.190	0.198	0.170	0.155
90	3 mm	0.193	0.217	0.224	0.238	0.203	0.188	0.208	0.217	0.186	0.170
30	5 mm	0.064	0.072	0.074	0.079	0.068	0.062	0.069	0.071	0.062	0.057
60	5 mm	0.157	0.177	0.183	0.195	0.165	0.153	0.170	0.177	0.152	0.139
90	5 mm	0.174	0.196	0.203	0.216	0.183	0.169	0.188	0.196	0.169	0.154

Table 4.6: Forward Solution for Normally Distributed Dipole Sheet Models (AE - Anterior Extension, OD - Diameter of Optic Disc)

	Solution at Each Electrode Location									
	1	2	3	4	5	6	7	8	9	10
Single	0.265	0.349	0.380	0.486	0.338	0.281	0.299	0.337	0.280	0.238

Table 4.7: Forward Solution for Single Dipole Model

between 8.092 and 53.94. The ten ocular standing potential models having the lowest scores were selected for further analysis. Table 4.11 lists the top ten models: the models are listed in order from the one having the lowest score to the highest. The results for the single dipole model are also included. Table 4.12 lists the mean, standard deviation, and accuracy scores for these ten models as well as the single dipole model.

It can be seen that the standard deviations of the best ocular standing potential models are all very close, indicating that these models will all result in a similar

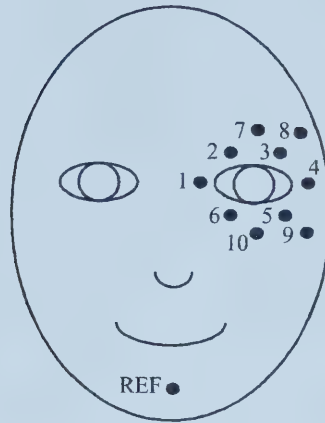


Figure 4.13: Electrode Locations

amount of variation in surface potential values. The mean values are also very close, meaning that each model will generate surface potential strengths that will be close to those of the other models.

Eight of the best ten models have dipole strengths that are linearly decreasing. Half of the models have an anterior extension of 60 degrees and half have an anterior extension of 90 degrees. These results indicate that the anatomical source of the EOG is likely to be partly due to incident light on the retina, and that this behaviour can be modeled with a linear distribution of dipole strength. It also appears that the portion of the pigment epithelium that gives rise to the standing potential extends 60 to 90 degrees forward of the back of the eye, measured with respect to the posterior-anterior axis.

All possible optic disc diameters are present in the best ten ocular standing

potential models. This indicates that the presence or absence of the optic disc region in the model does not significantly affect the surface potentials.

The results also seem to indicate that the single dipole source is a good approximation of the ocular standing potential, since the values of its metrics are closer to those of the recorded surface potentials than the distributed models are. This result contradicts the hypothesis that the ocular standing potential would be most accurately modeled with a distributed model. For this reason, further analysis of the ten best distributed models and the single dipole model was conducted, and is discussed in the next chapter.

Model Type	AE	OD	Mean	SD	Model Type	AE	OD	Mean	SD
Uniform	30	0 mm	0.174	0.093	Linear	30	0 mm	0.109	0.093
Uniform	60	0 mm	1.061	0.103	Linear	60	0 mm	0.357	0.101
Uniform	90	0 mm	1.997	0.106	Linear	90	0 mm	0.400	0.101
Uniform	30	1 mm	0.172	0.093	Linear	30	1 mm	0.108	0.093
Uniform	60	1 mm	1.059	0.103	Linear	60	1 mm	0.357	0.101
Uniform	90	1 mm	1.995	0.106	Linear	90	1 mm	0.399	0.101
Uniform	30	3 mm	0.153	0.094	Linear	30	3 mm	0.096	0.094
Uniform	60	3 mm	1.040	0.103	Linear	60	3 mm	0.344	0.101
Uniform	90	3 mm	1.976	0.106	Linear	90	3 mm	0.388	0.102
Uniform	30	5 mm	0.126	0.095	Linear	30	5 mm	0.079	0.095
Uniform	60	5 mm	0.997	0.104	Linear	60	5 mm	0.319	0.102
Uniform	90	5 mm	1.933	0.107	Linear	90	5 mm	0.363	0.103
Quadratic	30	0 mm	0.148	0.093	Normal	30	0 mm	0.094	0.093
Quadratic	60	0 mm	0.566	0.101	Normal	60	0 mm	0.197	0.099
Quadratic	90	0 mm	0.649	0.102	Normal	90	0 mm	0.215	0.099
Quadratic	30	1 mm	0.147	0.093	Normal	30	1 mm	0.093	0.093
Quadratic	60	1 mm	0.565	0.101	Normal	60	1 mm	0.356	0.101
Quadratic	90	1 mm	0.647	0.102	Normal	90	1 mm	0.214	0.099
Quadratic	30	3 mm	0.140	0.093	Normal	30	3 mm	0.083	0.094
Quadratic	60	3 mm	0.558	0.101	Normal	60	3 mm	0.186	0.099
Quadratic	90	3 mm	0.631	0.102	Normal	90	3 mm	0.204	0.100
Quadratic	30	5 mm	0.126	0.092	Normal	30	5 mm	0.068	0.095
Quadratic	60	5 mm	0.538	0.101	Normal	60	5 mm	0.167	0.100
Quadratic	90	5 mm	0.597	0.103	Normal	90	5 mm	0.185	0.101
Single			0.325	0.219					

Table 4.8: Forward Solution Metrics for Ocular Standing Potential Models (AE - Anterior Extension, OD - Diameter of Optic Disc, SD - Standard Deviation)

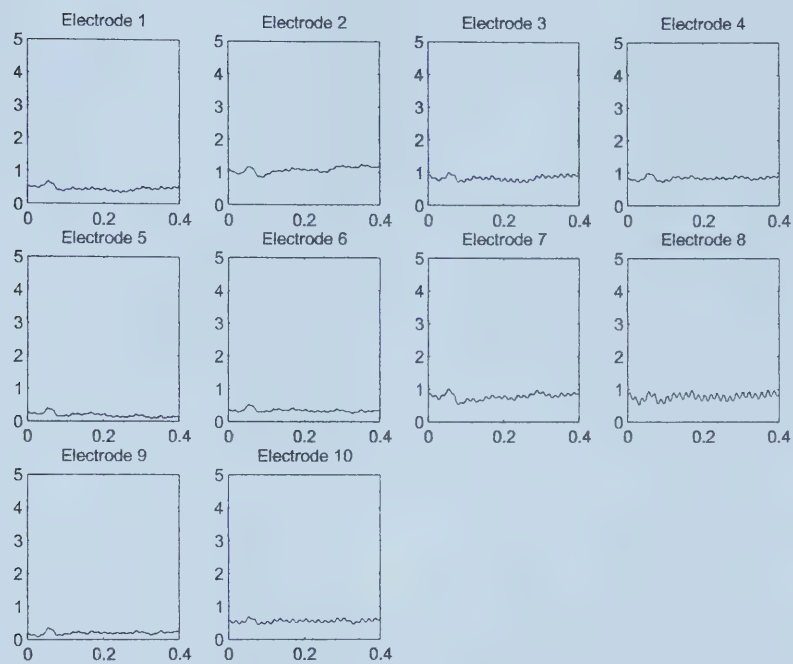


Figure 4.14: Filtered Signal from Each Electrode vs. Time (s)

Data Set	Electrode Number									
	1	2	3	4	5	6	7	8	9	10
1	0.246	0.119	0.668	0.891	0.568	0.621	0.307	0.230	0.388	0.138
2	0.508	0.370	0.426	0.954	0.901	0.387	0.320	0.785	0.398	1.072
3	0.453	1.101	0.827	0.865	0.226	0.318	0.766	0.806	0.214	0.560
4	0.504	0.940	0.788	0.836	0.083	0.318	0.677	0.617	0.137	0.435
5	0.547	0.873	0.675	0.754	0.371	0.318	0.724	0.745	0.185	0.277
6	0.432	1.068	0.788	0.897	0.292	0.107	0.892	0.802	0.253	0.556
7	0.212	0.299	0.213	0.303	0.659	0.482	0.135	0.871	0.115	0.084
8	0.212	0.299	0.213	0.303	0.817	0.492	0.135	0.871	0.051	0.150
9	0.212	0.032	0.287	0.555	0.717	0.664	0.135	0.511	0.060	0.017
10	0.212	0.299	0.213	0.303	0.659	0.482	0.135	0.871	0.115	0.084

Table 4.9: Mean of Recorded Potentials at Each Electrode

Data Set	SD (of the ten signal values)	Mean
1	0.616	0.418
2	0.465	0.612
3	0.495	0.614
4	0.547	0.534
5	0.441	0.547
6	0.535	0.609
7	0.764	0.337
8	0.803	0.354
9	0.847	0.319
10	0.764	0.337

Table 4.10: Standard Deviation and Mean of Recorded Data Sets

Model Type	AE	OD	Data Set				
			1	2	3	4	5
Linear	60	5 mm	0.793	0.955	0.966	0.754	0.844
Linear	60	3 mm	0.764	1.033	0.906	0.751	0.837
Normal	60	1 mm	0.753	1.069	0.879	0.753	0.836
Linear	60	n/a	0.752	1.073	0.876	0.753	0.836
Linear	60	1 mm	0.752	1.073	0.876	0.753	0.836
Linear	90	5 mm	0.745	1.093	0.861	0.754	0.836
Linear	90	3 mm	0.729	1.171	0.806	0.766	0.842
Linear	90	1 mm	0.724	1.208	0.782	0.774	0.847
Linear	90	n/a	0.724	1.212	0.779	0.775	0.847
Normal	90	n/a	0.981	0.624	1.245	0.850	0.949
Single			0.708	0.991	0.934	0.757	0.844

Model Type	AE	OD	Data Set				
			6	7	8	9	10
Linear	60	5 mm	0.795	0.696	0.786	0.747	0.754
Linear	60	3 mm	0.823	0.723	0.789	0.763	0.751
Normal	60	1 mm	0.839	0.738	0.793	0.773	0.753
Linear	60	n/a	0.841	0.740	0.794	0.774	0.753
Linear	60	1 mm	0.841	0.740	0.794	0.774	0.753
Linear	90	5 mm	0.850	0.749	0.796	0.780	0.754
Linear	90	3 mm	0.889	0.788	0.812	0.808	0.766
Linear	90	1 mm	0.909	0.808	0.823	0.823	0.774
Linear	90	n/a	0.911	0.810	0.824	0.825	0.775
Normal	90	n/a	0.755	0.676	0.858	0.773	0.850
Single			0.808	0.704	0.749	0.753	0.757

Table 4.11: Sum of RMS Errors Between Predicted and Recorded Surface Potentials at Each Electrode Location (AE - Anterior Extension, OD - Optic Disc Diameter)

Model			Accuracy			
Type	AE	OD	Mean	SD	Score	
Linear	60	5 mm	0.319	0.102	8.092	
Linear	60	3 mm	0.344	0.101	8.141	
Normal	60	1 mm	0.356	0.101	8.186	
Linear	60	n/a	0.357	0.101	8.192	
Linear	60	1 mm	0.357	0.101	8.192	
Linear	90	5 mm	0.363	0.103	8.218	
Linear	90	3 mm	0.388	0.102	8.376	
Linear	90	1 mm	0.399	0.101	8.471	
Linear	90	n/a	0.400	0.101	8.482	
Normal	90	n/a	0.215	0.099	8.561	
Single			0.325	0.219	8.005	

Table 4.12: Mean, Standard Deviation, and Accuracy Score of Best Ocular Standing Potential Models (AE - Anterior Extension, OD - Optic Disc Diameter)

Chapter 5

Electrode Placement

This chapter discusses biopotential electrodes for use in the ocular prosthesis system, and then describes the results of simulations that were performed to determine the optimum placement of electrodes in the clinical ocular prosthesis system.

5.1 Biopotential Electrodes

Biopotentials are due to electrochemical activity within the body. Current flow in the body is facilitated by the movement of ions, not electrons. Therefore, detecting biopotential signals such as the EOG involves transducing the ionic currents into electrical currents. This transduction is done using electrodes that consist of electrical conductors contacting an aqueous ionic solution. This is shown in Figure 5.1 for an imaginary biopotential electrode consisting of metal X in contact with an electrolyte containing X^+ ions.

The electrode can be oxidized to form a cation and a free electron, as shown in Equation 5.1.



In general, this reaction is reversible. In steady-state, the rate of oxidation reactions equals the rate of reduction reactions, and no current flows.

When the electrode is introduced into the electrolyte, the relative concentrations

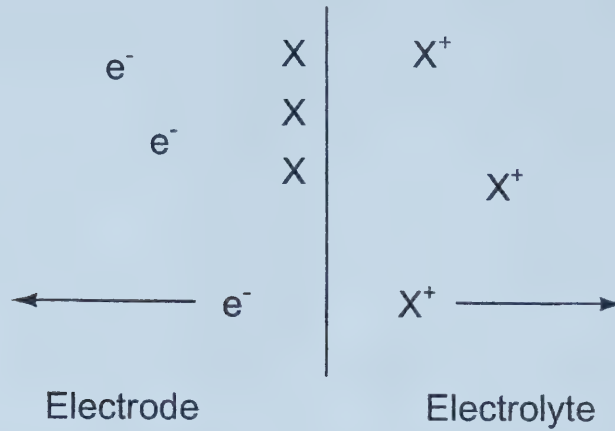
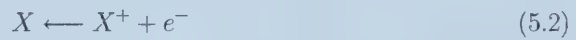


Figure 5.1: Biopotential Electrode

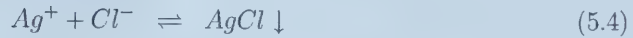
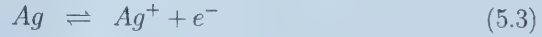
of ions in the electrolyte determine whether oxidation or reduction reactions predominate. For example, if the electrolyte has a higher concentration of cations than anions, reduction reactions will predominate, as given in Equation 5.2. The resulting flow of electrons towards the electrolyte results in current flow from the electrolyte to the electrode, which attracts cations towards the electrode-electrolyte interface.



This movement of ions causes a higher concentration of cations in the vicinity of the electrode compared to the bulk of the electrolyte. The potential difference between these regions of electrolyte is known as the half-cell potential.

Electrodes that are applied externally use an electrolytic gel to couple the electrode to the skin. The electrodes that are commonly used for EOG detection are dis-

posable, self-adhesive silver/silver chloride electrodes [53]. The reduction-oxidation reactions that govern the behaviour of the silver/silver chloride electrode are given in Equations 5.3 and 5.4. The reaction in Equation 5.4 immediately follows the reaction in Equation 5.3 in the oxidation reaction, and vice versa in the reduction reaction.



Silver/silver chloride electrodes are commonly used as biopotential electrodes because they have a low half-cell potential. These electrodes are almost perfectly non-polarizable, meaning that the activation energy required to allow the current to flow through the electrode is very low. This means that the half-cell potential of a silver/silver chloride electrode is essentially the same whether or not current is flowing. This low, stable half-cell potential is a desirable characteristic in a biopotential electrode because it results in a more stable, less noisy recording of the biopotential.

Figure 5.2 shows the equivalent circuit model of a biopotential electrode that is coupled to the skin via an electrolyte gel. E_{hc} is the half-cell potential of the silver/silver chloride electrode. E_{se} is the potential difference across the stratum corneum, which is the outermost layer of the epidermis and is semi-permeable to ions. The parallel combination of C_d and R_d represents the impedance of the electrode-electrolyte interface. R_s represents the impedance of the electrolyte and the impedance associated with interface effects of the electrolyte gel. The parallel combination of C_e and R_e is the impedance of the epidermis, and R_u is the impedance

of the dermis and subcutaneous layers.

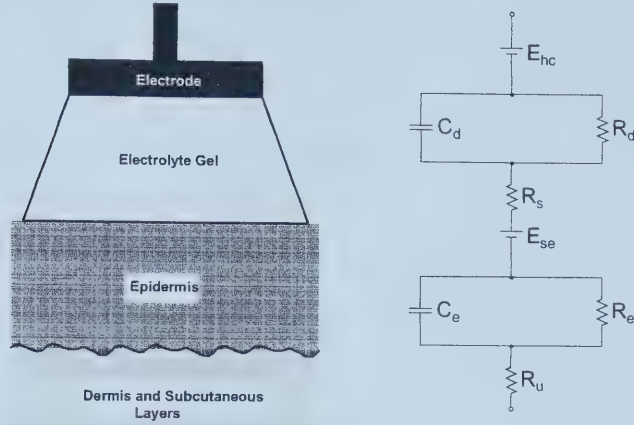


Figure 5.2: Biopotential Electrode Equivalent Circuit Model [17]

Electrode movement can disturb the ionic concentrations at the electrode-electrolyte and electrolyte-skin interfaces. In non-polarizable electrodes such as the silver/silver chloride electrode, a change in ionic concentration at the electrolyte-skin interface can result in a change in the potential E_{se} associated with this interface. This results in a DC offset in the recorded signal that is known as a motion artifact. Polarizable electrodes can suffer from motion artifacts from both the electrode-electrolyte and electrolyte-skin interfaces.

The severity of motion artifacts can be reduced by providing a conduction path through the stratum corneum. This is typically done by abrading the skin with fine sandpaper or an isopropyl alcohol wipe to remove the few cells that comprise this layer of the skin [17]. This procedure also shorts out the parallel combination of R_e

and C_e , reducing the impedance of the electrode-skin connection.

For internal EOG measurement, a platinum needle or wire electrode would most likely be used. Internal electrodes do not require a coupling fluid because bodily fluids serve the same purpose. Platinum electrodes are polarizable, so it would be important to securely fix them in place to reduce motion artifacts. Also, platinum is relatively inert, making these electrodes more biocompatible than silver/silver chloride electrodes. Internal electrodes would be surgically inserted, and would remain in place for years at a time. There are many clinical examples of long-term use of electrodes. For example, they are used in cardiac pacing to detect the electrocardiogram of the heart. As early as 1975, research had shown that fine wire electrodes could be left in skeletal muscle for years without adverse effects [54].

Electrodes that are left in the body for extended periods can cause an inflammatory reaction over time in the surrounding tissue. This can cause fibrous tissue to build up around the electrode. This new tissue is electrically different from the original tissue, and can cause problems with processing the signal of interest. One solution to this problem is to use steroid-eluting electrodes [55]. These electrodes have a semi-porous tip that is filled with a corticosteroid. Over time, the steroid slowly diffuses into the tissue surrounding the electrode and limits the inflammation that can result, preventing the build up of fibrous tissue.

5.2 Modeling Electrode Placement

The ten best models of the ocular standing potential as determined in Section 4.4 and the single dipole model were used to determine the optimum electrode config-

uration for the implanted ocular prosthesis. Approximately 2250 possible electrode locations in the vicinity of the eye were evaluated. These locations were all in the anterior superior quadrant (top front quadrant) of the head, on the same side as the natural eye. The locations were also 3 mm under the surface of the model, which approximately corresponded to the dermal or subcutaneous layers of the skin or the underlying muscle in the vicinity of the bony orbit [40]. This region corresponded to the feasible locations for the implanted electrodes in the clinical ocular prosthesis system. Locations in front of the eye itself were disregarded because electrodes cannot be implanted in this area.

The potential at each proposed electrode location due to the ocular standing potential model was calculated by solving the forward source localisation problem, as described in Section 4.2.

5.3 Results

Figures 5.3 and 5.4 show the strength of the surface potentials at each proposed electrode location for the ten best standing potential models as well as the single dipole source. The colour of the plot indicates the strength of the potential, with the red areas being the strongest. The strength values are displayed raised to the power 2.2 in order to increase the contrast. The top and bottom edges of each plot are closest to the top and bottom of the head, respectively. The left edge of each plot corresponds to the side of the head, and the right edge corresponds to the nose. The black region in the centre of each figure corresponds to possible electrode locations in front of the eye, which were disregarded. It can be seen that the region of greatest

signal strength is the same in all cases, namely the area superior and lateral to the natural eye.

The single dipole model gives rise to a region of high surface potential strength just lateral to the outer canthus of the eye. It can be determined by comparing the predicted surface potential values for the single dipole source with the recorded surface potentials that the contour produced by the single dipole source is inaccurate. The area of strong potentials predicted near the outer canthus of the eye does not occur in the recorded surface potentials. The contours produced by the distributed models are more accurate in that they exhibit the diffusion of signal strength that is seen in the recorded surface potentials. Therefore, the single dipole source is not an accurate model of the ocular standing potential.

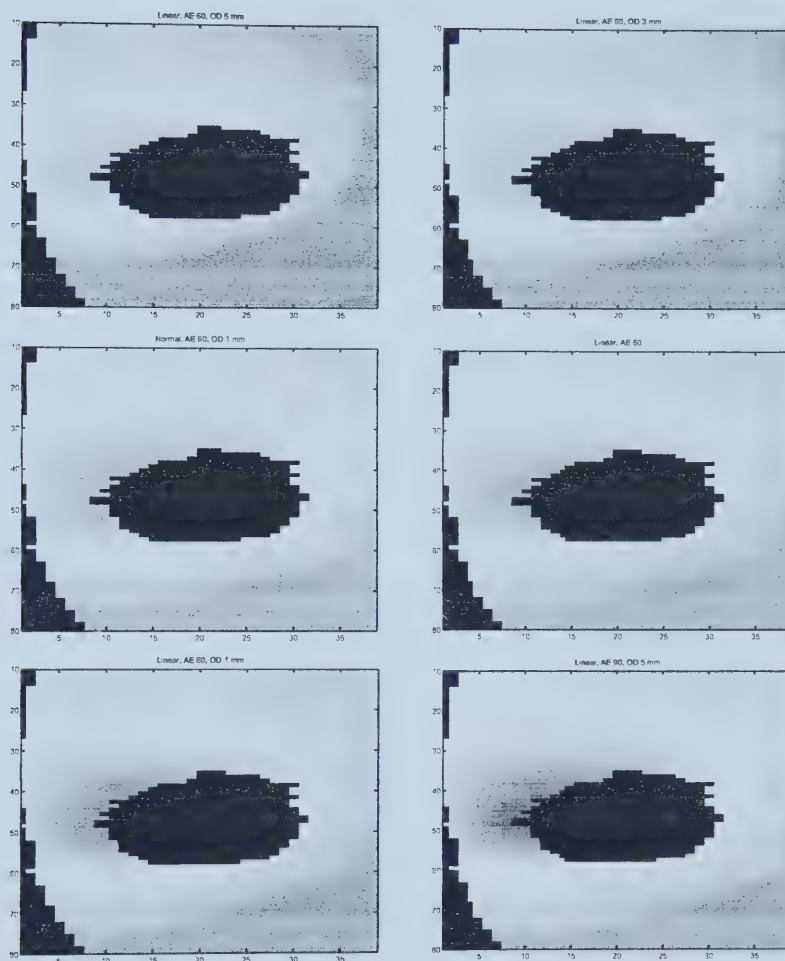


Figure 5.3: Strength of Surface Potentials Generated by Ocular Standing Potential Models

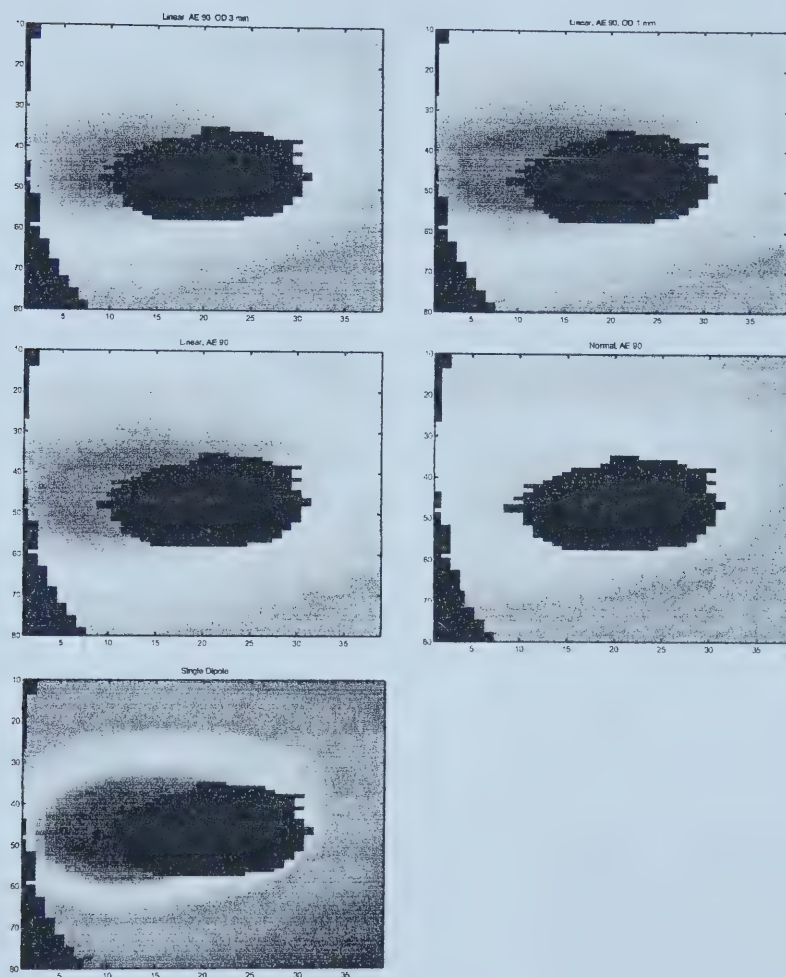


Figure 5.4: Strength of Surface Potentials Generated by Ocular Standing Potential Models (continued)

Chapter 6

Conclusions

6.1 Eye Movement Detection Method

The electrooculogram method appears to have good potential for detecting the position of the natural eye in the ocular prosthesis system. This method provides several advantages over other eye movement detection methods. The location of the EOG in the vicinity of the natural eye provides surgeons with easy electrode placement. Once the subcutaneous electrodes are implanted they will not be noticeable to the observer, providing the patient with a good cosmetic result. Also, a control signal can be extracted from the EOG using limited processing power of the ocular prosthesis, and the EOG method is able to satisfy the design requirements given in Section 1.2.

6.2 Signal Processing

A biopotential amplifier should be used to amplify the EOG signal detected from the implanted electrodes. This type of amplifier is advantageous because it provides good common mode noise rejection and gain as well as patient protection. It was found that a 5th-order Chebyshev type II filter with a cut-off frequency of 10 Hz was suitable for de-noising the EOG signal following amplification.

Minimizing lag is important to the performance of the ocular prosthesis system

because the observer may notice a lag of greater than 1 ms. Therefore, the lag introduced by the Chebyshev filter should be compensated in the control system that governs the movement of the ocular prosthesis. The control system should also be designed to meet the rise time and steady state ripple specifications given in Section 1.2.

Several papers in the EOG signal processing literature reported that simple IIR filters introduced significant amounts of distortion to the duration and velocity of the EOG signal, as mentioned in Section 3.4. This was not found to be the case in the EOG signal processing experiments described in this thesis. It is likely that signal distortion would have been a problem if higher order IIR filters had been necessary to sufficiently de-noise the recorded EOG signal. However, the differential amplification techniques used to obtain the EOG signal provided enough common-mode noise rejection that higher order filters were unnecessary.

6.3 EOG Source Model

The results presented in Chapter 4 indicate that the source of the EOG signal, the ocular standing potential, can be modeled as a dipole sheet of linearly decreasing dipole strength located at the back of the eye that extends forward 60 to 90 degrees from the anterior-posterior axis of the eye. This suggests that the pigment epithelium is the source of the standing potential, and that incident light contributes to the standing potential.

The optic disc region does not appear to contribute significantly to the model. This is most likely because the area of the optic disc region is small compared to the

area of the entire dipole sheet.

The single dipole model was found to produce an inaccurate surface potential distribution compared with the recorded surface potential distribution.

6.3.1 Error Analysis

There was significant error between the predicted and measured surface potentials, even for the best ocular standing potential models. There are several possible sources of this error.

The most likely source of error is the head model. Although the head model that was used was much more accurate than the standard concentric shell model, it was still an approximation of the anatomical data from which it was developed. In particular, the bony regions were much thicker in the head model than was the case in the human skull. This would have caused errors in the predicted surface potentials because the conductivity of bone is much lower than that of the surrounding tissue. As such, bone has a large attenuating effect on the distribution of signal strength throughout the head.

Another possible source of error is electrode movement. It is difficult to fix electrodes to the face near the nose because of the shape of the face. Either weak contact between the skin and the electrolyte contained within the electrode or electrode movement could have generated errors in the collected data.

Also, it was only possible to collect experimental data from one person during the data collection phase of this project. This may have resulted in errors in the determination of the most accurate ocular standing potential models, since this one individual may not be representative of the majority of the population.

6.4 Electrode Configuration

It was shown in Chapter 5 that the ten best ocular standing potential models all give rise to a similar contour of surface potential strengths.

Based on the contour plots shown in Chapter 5, the best locations for the implanted electrodes would be above the eye. One electrode should be placed 5 mm above the outer canthus of the eye. The second electrode should be placed 5 mm above the inner canthus and 5 mm towards the midline of the eye. This electrode placement provides a good compromise between locating the electrodes where the surface potentials are strongest, and having sufficient separation between the electrodes to perform a bipolar recording. These areas are also readily accessible to the surgeons who will perform the implantation procedure, and pose little risk to the remaining natural eye. The reference electrode can be implanted at any location that is indifferent to the EOG. A suitable location for the reference electrode would be under the skin in the centre of the forehead. If the electrodes are securely fixed to the surrounding tissues, movement of the electrodes relative to the orbit position is unlikely. This will greatly reduce motion artifacts in the EOG signal.

Using two electrodes allows the bipolar nature of the EOG recording to be retained. This configuration is highly desirable as it allows for common mode rejection of noise in the EOG signal, which results in a much cleaner EOG recording. The difference in signal strength between the two suggested electrode locations does not pose a problem, since the ocular prosthesis system can be calibrated to determine the correct eye position from a given bipolar EOG signal.

Steroid-eluting electrodes should be used in order to limit the build-up of fibrous

tissue around the electrode, which would attenuate the surface potentials resulting from the ocular standing potential.

6.5 Future work

Future work on this project could include repeating the experiments with a more accurately segmented head model and verifying that the ocular standing potential model is valid during eye movement, once more computing resources are available. Also, experimental data should be collected from a larger number of experimental subjects, and this data should be used to verify the results obtained in this thesis.

The conclusions of this thesis should also be applied to the development of the clinical ocular prosthesis system.

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Appendix A: Ethical Approval

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UNIVERSITY OF ALBERTA HEALTH SCIENCES FACULTIES,
CAPITAL HEALTH AUTHORITY, AND CARITAS HEALTH GROUP

HEALTH RESEARCH ETHICS APPROVAL

Date: October 2002

Name of Applicant: Ms. Cheryl Card & Dr. Zoltan Koles

Organization: University of Alberta

Department: Electrical Engineering

Name of Co-applicant: Dr. Al Cook

Organization: University of Alberta

Department: Rehabilitation Medicine

Project Title: Modelling the source of the electrooculogram signal for controlling a prosthetic eye.

The Health Research Ethics Board (HREB) has reviewed the protocol for this project and found it to be acceptable within the limitations of human experimentation. The HREB has also reviewed and approved the subject information letter and consent form.

The deliberations of the HREB included all elements described in Section 50 of the *Health Information Act*, and found the study to be in compliance with all the applicable requirements of the Act.

The approval for the study as presented is valid for one year. It may be extended following completion of the yearly report form. Any proposed changes to the study must be submitted to the Health Research Ethics Board for approval. Written notification must be sent to the HREB when the project is complete or terminated.



Dr. Sharon Warren

Chair of the Health Research Ethics Board (B: Health Research)

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